IN THE UNITED STATES DISTRICT COURT
FOR THE DISTRICT OF DELAWARE

ENERGY TRANSPORTATION GROUP, INC.,	)	
	)	C.A. No. ______________
Plaintiff,	)	
	)	
v.	)	
	)	
SIEMENS HEARING INSTRUMENTS, INC. and SIEMENS AUDIOLOGISCHE TECHNIK GMBH,	)	**JURY TRIAL DEMANDED**
	)	
Defendants.	)	

COMPLAINT AND JURY DEMAND

1. Plaintiff ENERGY TRANSPORTATION GROUP, INC. ("ETG" or "Plaintiff"), by and through its attorneys, hereby demands a jury trial and complains of Defendant SIEMENS HEARING INSTRUMENTS, INC. ("Siemens Hearing Instruments") and SIEMENS AUDIOLOGISCHE TECHNIK GmbH ("Siemens Audiologische Technik GmbH" or "Siemens AT GmbH") (collectively, "Defendants") as follows:

NATURE OF THE ACTION

2. This is an action for patent infringement arising under the patent laws of the United States, 35 U.S.C. §§ 271, *et seq.* to enjoin and obtain damages resulting from Defendants' unauthorized manufacture, use, sale, offer to sell and/or importation into the United States for subsequent use or sale of products, methods, processes, services and/or systems that infringe one or more claims of United States Patent No. 4,731,850 (the "'850 Patent") entitled "Programmable Digital Hearing Aid System" and one or more claims of United States Patent No. 4,879,749 (the "'749 Patent") entitled "Host Controller for Programmable Digital Hearing

Aid System." A copy of the '850 Patent is attached as Exhibit A and a copy of the '749 Patent is attached as Exhibit B. Plaintiff seeks injunctive relief to prevent Defendants from continuing to infringe Plaintiff's '850 Patent and the '749 Patent. In addition, Plaintiff seeks a recovery of monetary damages resulting from Defendants' past infringement of the '850 Patent and the '749 Patent.

3. This action for patent infringement involves Defendants' manufacture, use, sale, offer for sale, and/or importation into the United States of infringing products, methods, processes, services and systems that are primarily used or primarily adapted for use of or in a programmable digital hearing aid device.

4. Plaintiff has been irreparably harmed by Defendants' infringement of its valuable patent rights. Moreover, Defendants' unauthorized, infringing use of Plaintiff's patented systems and methods has threatened the value of this intellectual property because Defendants' conduct results in Plaintiff's loss of its lawful patent rights to exclude others from making, using, selling, offering to sell and/or importing the patented inventions.

THE PARTIES

5. Plaintiff ETG is a Delaware corporation organized and existing under the laws of Delaware and having its principal place of business at 654 Madison Avenue, Suite 1705, New York, NY 10021.

6. Plaintiff ETG is the lawful assignee of all right, title and interest in and to the '850 Patent and the '749 Patent. The '850 Patent was lawfully issued on March 15, 1988 in the name of Dr. Harry Levitt, Richard S. Dugot and Kenneth W. Kopper, as the named inventors. The '749 Patent was lawfully issued on November 7, 1989 in the name of Dr. Harry Levitt, Richard S. Dugot and Kenneth W. Kopper as the named inventors. Plaintiff ETG's predecessors,

Audimax Corporation and Audimax, Inc., were involved in developing innovation for use in the hearing instrument industry.

7. Siemens is a corporation organized and existing under the laws of Delaware, and maintains a place of business at 10 Constitution Avenue, P.O. Box 1397, Piscataway, New Jersey 08855.

8. Siemens GmbH is a corporation organized and existing under the laws of Germany, and maintains a place of business at Gebbertstrasse 125, 91058 Erlangen, Germany.

JURISDICTION AND VENUE

9. This Court has jurisdiction over the subject matter of this patent infringement action pursuant to 28 U.S.C. §§ 1331 and 1338(a). The Court has personal jurisdiction over the Defendants in that each has committed acts within Delaware and this judicial district giving rise to this action and each of the Defendants has established minimum contacts with the forum such that the exercise of jurisdiction over each of the Defendants would not offend traditional notions of fair play and substantial justice.

10. ETG is a Delaware corporation. Venue is proper in this district pursuant to 28 U.S.C. §§ 1391(b), 1391(c), and 1400(b) for at least the reasons that the Defendants each have committed acts within this judicial district giving rise to this action and does business in this district, including sales, and providing service and/or support to their respective customers in this district.

COUNT I

(Patent Infringement of United States Patent No. 4,731,850)

11. Paragraphs 1 through 10 are incorporated by reference as if fully restated herein.

12. Defendants make, use, sell, offer to sell and/or import into the United States for subsequent sale or use products, services, methods or processes that infringe directly and/or indirectly, which employ systems, components and/or steps that make use of other systems or processes that infringe directly and/or indirectly or which are made according to a patented process, one or more of the claims of the '850 Patent.

13. Defendants have been and continue infringing one or more of the claims of the '850 Patent through the aforesaid acts, and will continue to do so unless enjoined by this Court. Defendants' wrongful conduct has caused Plaintiff to suffer irreparable harm resulting from the loss of its lawful patent rights to exclude others from making, using, selling, offering to sell and importing the patented inventions.

14. Plaintiff is entitled to recover damages adequate to compensate for the infringement.

COUNT II

(Willful Patent Infringement of United States Patent No. 4,731,850)

15. Paragraphs 1 through 14 are incorporated by reference as if fully restated herein.

16. Defendants' infringement has been willful, deliberate and with knowledge of Plaintiff's rights under the '850 Patent, and unless Defendants are enjoined by this Court, such acts of willful infringement by Defendants will continue. Therefore, Plaintiff is without adequate remedy at law. ETG is entitled to recover damages adequate to compensate for the infringement of the '850 Patent, as well as additional damages for willful infringement including increased

damages under 35 U.S.C. § 284 and to attorneys' fees and costs incurred in prosecuting this action under 35 U.S.C. § 285.

COUNT III

(Patent Infringement of United States Patent No. 4,879,749)

17. Paragraphs 1 through 16 are incorporated by reference as if fully restated herein.

18. Defendants make, use, sell, offer to sell and/or import into the United States for subsequent sale or use products, services, methods or processes that infringe directly and/or indirectly, which employ systems, components and/or steps that make use of other systems or processes that infringe directly and/or indirectly or which are made according to a patented process, one or more of the claims of the '749 Patent.

19. Defendants have been and continue infringing one or more of the claims of the '749 Patent through the aforesaid acts, and will continue to do so unless enjoined by this Court. Defendants' wrongful conduct has caused Plaintiff to suffer irreparable harm resulting from the loss of its lawful patent rights to exclude others from making, using, selling, offering to sell and importing the patented inventions.

20. Plaintiff is entitled to recover damages adequate to compensate for the infringement.

COUNT IV

(Willful Patent Infringement of United States Patent No. 4,879,749)

21. Paragraphs 1 through 20 are incorporated by reference as if fully restated herein.

22. Defendants' infringement has been willful, deliberate and with knowledge of Plaintiff's rights under the '749 Patent, and unless Defendants are enjoined by this Court, such acts of willful infringement by Defendants will continue. Therefore, Plaintiff is without adequate

remedy at law. ETG is entitled to recover damages adequate to compensate for the infringement of the '749 Patent, as well as additional damages for willful infringement including increased damages under 35 U.S.C. § 284 and to attorneys' fees and costs incurred in prosecuting this action under 35 U.S.C. § 285.

JURY DEMAND

23. Plaintiff demands a trial by jury of all matters to which it is entitled to trial by jury pursuant to FED. R. CIV. P. 38.

PRAYER FOR RELIEF

WHEREFORE, Plaintiff prays for judgment against Defendants, granting Plaintiff the following relief:

A. That this Court adjudge and decree that the '850 Patent is valid and enforceable against Defendants;

B. That this Court adjudge and decree that the '749 Patent is valid and enforceable against Defendants;

C. That this Court adjudge and decree that Defendants have infringed and continue to infringe the '850 Patent;

D. That this Court adjudge and decree that Defendants have infringed and continue to infringe the '749 Patent;

E. That this Court order an accounting of all damages sustained by ETG as the result of the acts of infringement by each Defendant;

F. That this Court enter an award to Plaintiff of such damages as it shall prove at trial against Defendants that are adequate to compensate Plaintiff for said infringement, said damages to be no less than a reasonable royalty together with prejudgment interest and costs;

G. That this Court order an award to ETG of up to three times the amount of compensatory damages because of Defendants' willful infringement, and any enhanced damages provided by 35 U.S.C. § 284;

H. That this Court render a finding that this case is "exceptional" and award to ETG its costs and reasonable attorneys' fees, as provided by 35 U.S.C. § 285; and

I. That this Court grant to Plaintiff such other, further, and different relief as may be just and proper.

ASHBY & GEDDES

Steven J. Balick (I.D. #2114)
John G. Day (I.D. #2403)
Tiffany Geyer Lydon (I.D. #3950)
500 Delaware Avenue, 8th Floor
P.O. Box 1150
Wilmington, DE 19899
(302) 654-1888

Attorneys for Energy Transportation Group, Inc.

Of Counsel:

Marty Steinberg
HUNTON & WILLIAMS LLP
1111 Brickell Avenue
Suite 2500
Miami, Florida 33131
Telephone: (305) 810-2500
Facsimile: (305) 810-2460

Brian M. Buroker
Yisun Song
Robert L. Kinder, Jr.
Hunton & Williams LLP
1900 K Street, N.W.; Suite 1200
Washington, DC 20006
(T) (202) 955-1500
(F) (202) 778-2201

Maya M. Eckstein
HUNTON & WILLIAMS LLP
Riverfront Plaza, East Tower
951 East Byrd Street
Richmond, VA 23219-4074
Telephone: (804) 788-8788
Facsimile: (804) 343-4630

Dated: August 28, 2007
183636.1

EXHIBIT A

United States Patent [19]
Levitt et al.

[11] **Patent Number:** **4,731,850**
[45] **Date of Patent:** **Mar. 15, 1988**

[54] **PROGRAMMABLE DIGITAL HEARING AID SYSTEM**

[75] Inventors: **Harry Levitt**, Livingston, N.J.; **Richard S. Dugot**, New York, N.Y.; **Kenneth W. Kopper**, River Edge, N.J.

[73] Assignee: **Audimax, Inc.**, Hackensack, N.J.

[21] Appl. No.: **879,214**

[22] Filed: **Jun. 26, 1986**

[51] **Int. Cl.**[4] **H04B 15/00**
[52] **U.S. Cl.** **381/68.2**; 381/68.4; 381/94
[58] **Field of Search** 381/68.2, 68.4, 93, 381/94

[56] **References Cited**

U.S. PATENT DOCUMENTS

4,187,413	2/1980	Moser	381/68
4,425,481	1/1984	Mansgold	381/68.2
4,489,610	12/1984	Slavin	381/68
4,548,082	10/1985	Engebretson et al.	381/68
4,596,902	6/1986	Gilman	381/68.2
4,628,529	12/1986	Borth	381/68.2
4,630,305	12/1986	Borth	381/68.2

OTHER PUBLICATIONS

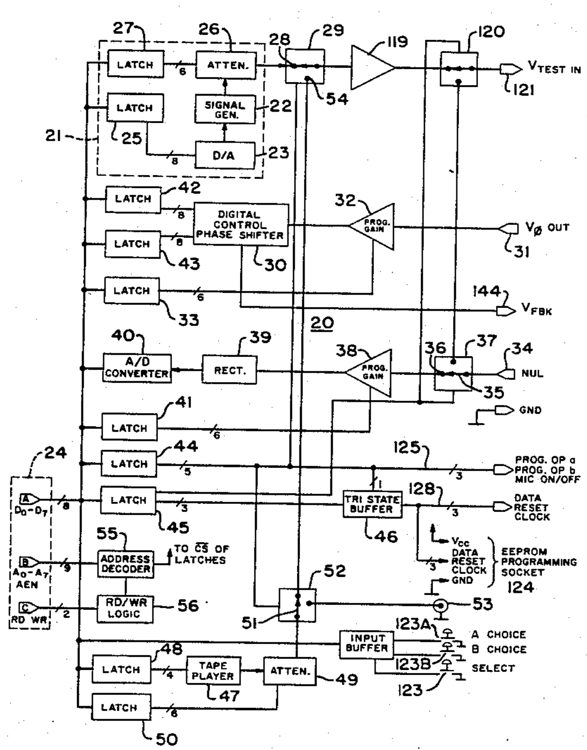
"Computer Applications in Audiology and Rehabilitation of the Hearing Impaired", by Harry Levitt, Journal of Communication Disorders 13, (1980), pp. 471–481.
E. G. & G. Reticon Technical Note No. 105, "A Tapped Analog Delay for Sampled Data Signal Processing", pp. 7–15–7–17, 7–25 and 7–26.

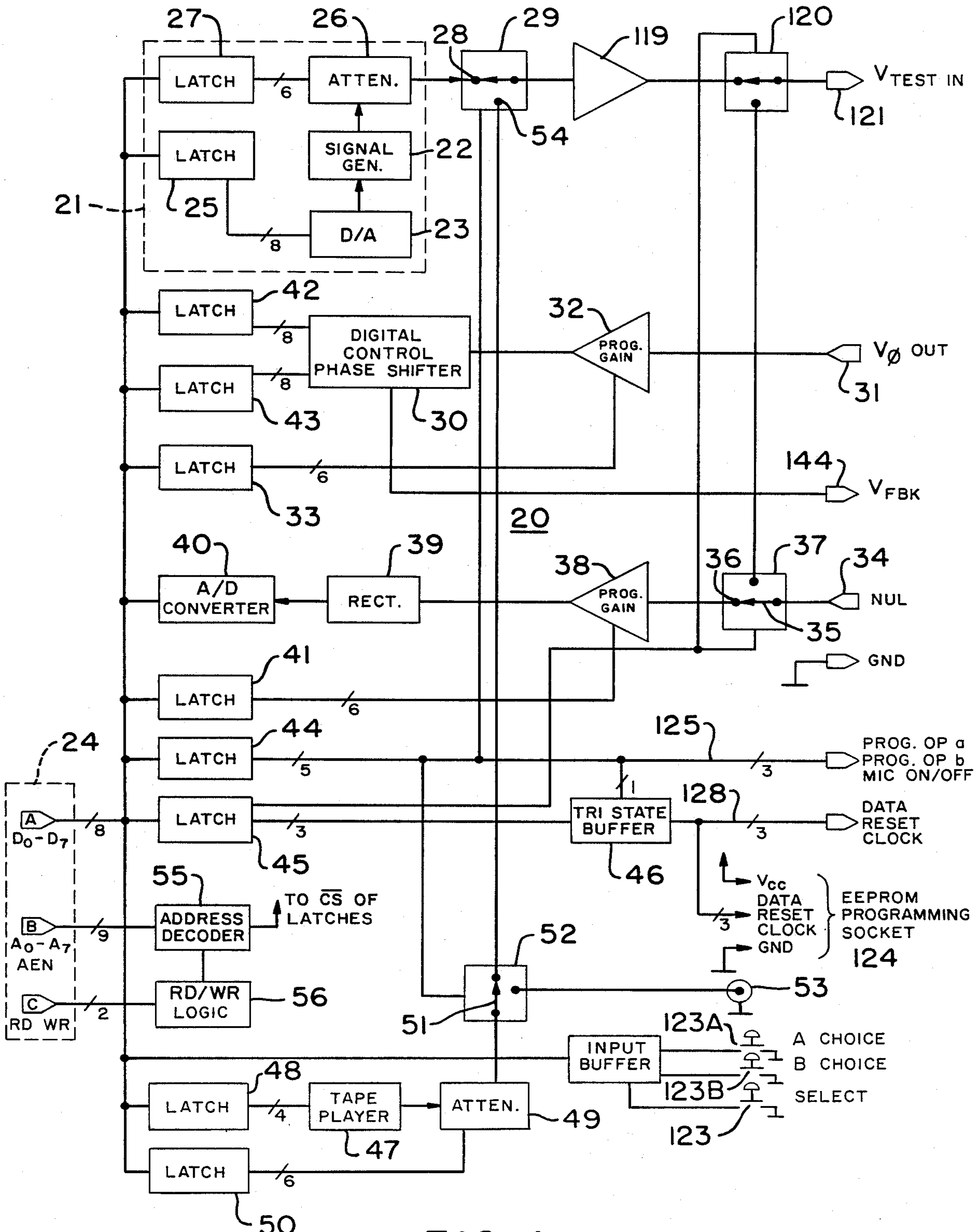
Primary Examiner—Jin F. Ng
Assistant Examiner—L. C. Schroeder
Attorney, Agent, or Firm—Brumbaugh, Graves, Donohue & Raymond

[57] **ABSTRACT**

A hearing aid system comprises a hearing aid that is programmable so as to have optimum electro-acoustic characteristics for the patient and acoustic environment in which it is used. Selected optimum parameter values are programmed into an electronically erasable, programmable read only memory (EEPROM) which supplies coefficients to a programmable filter and amplitude limiter in the hearing aid so as to cause the hearing aid to adjust automatically to the optimum set of parameter values for the speech level, room reverberation, and type of background noise then obtaining. The programmable filter may be a digital equivalent of a tapped delay line in which each delayed sample is multiplied by a weighting coefficient and the sum of the weighted samples generates a desired electro-acoustic characteristic. Alternatively, the programmable filter may be a tapped analog delay line in which the sum of the weighted outputs of the taps generates the desired characteristics. Acoustical feedback is reduced by an electrical feedback path in the hearing aid which is matched in both amplitude and phase to the acoustic feedback path, the two feedback signals being subtracted so as to cancel each other.

19 Claims, 5 Drawing Figures





LATCH
ATTEN.
SIGNAL GEN.
D/A
DIGITAL CONTROL PHASE SHIFTER
PROG. GAIN
A/D CONVERTER
RECT.
TRI STATE BUFFER
ADDRESS DECODER
RD/WR LOGIC
TO CS OF LATCHES
INPUT BUFFER
TAPE PLAYER
V_{TEST} IN
$V_Ø$ OUT
V_{FBK}
NUL
GND
PROG. OP a
PROG. OP b
MIC ON/OFF
DATA
RESET
CLOCK
V_{cc}
EEPROM PROGRAMMING SOCKET
A CHOICE
B CHOICE
SELECT
D_0-D_7
A_0-A_7
AEN
RD WR
20

FIG. 1

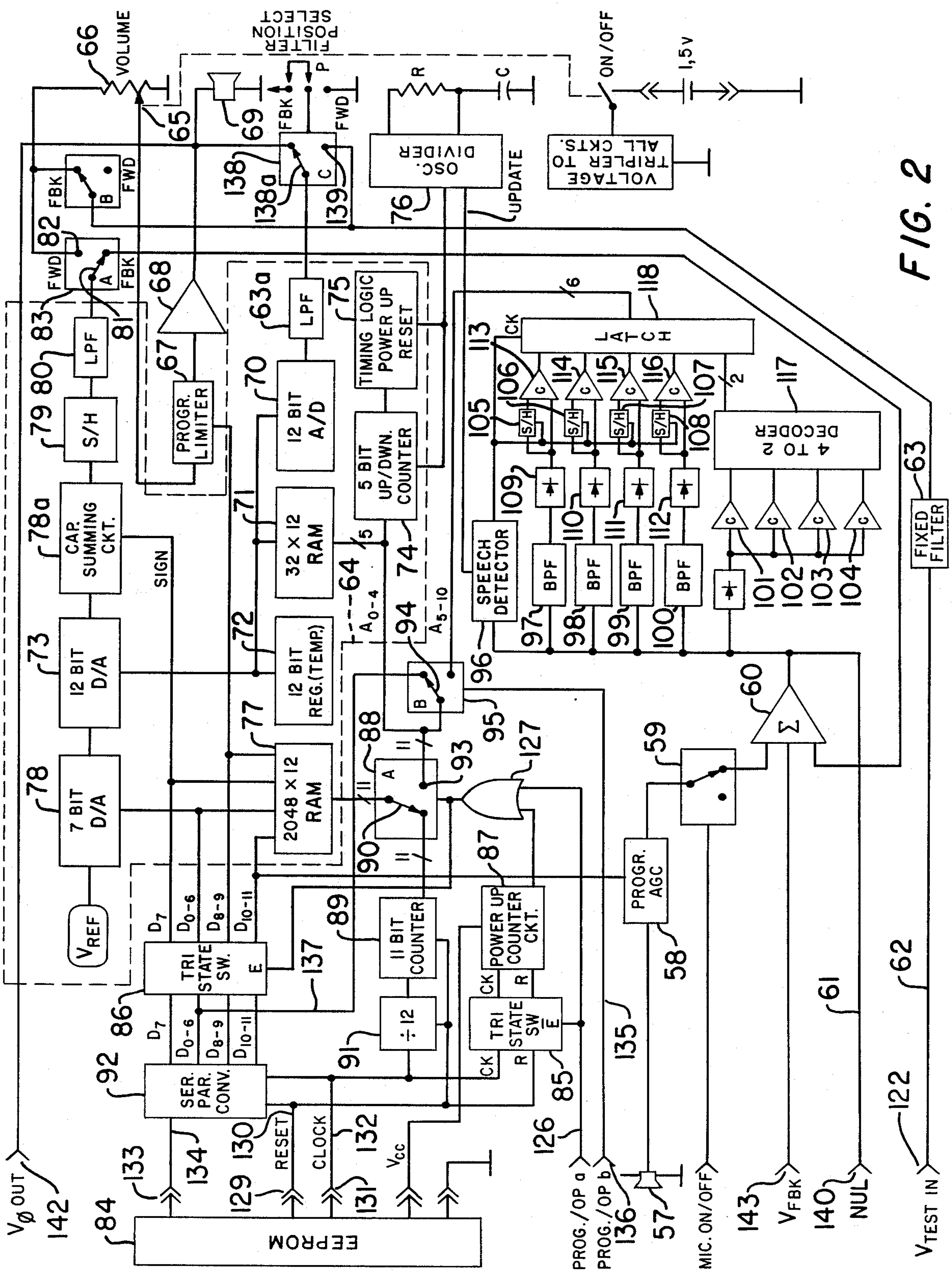
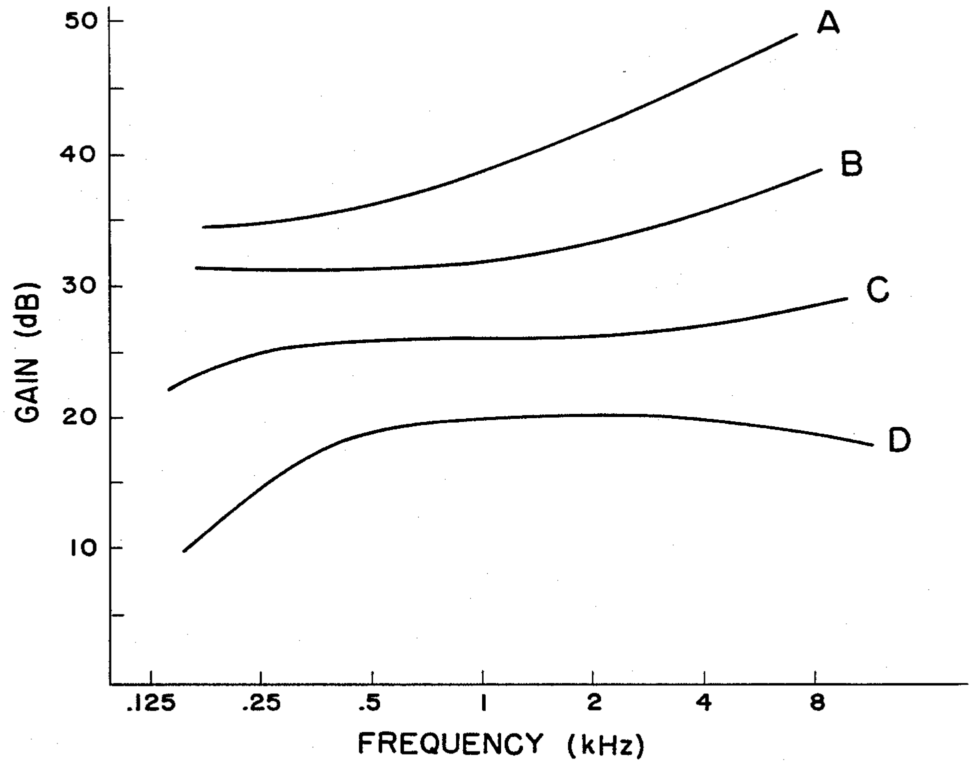
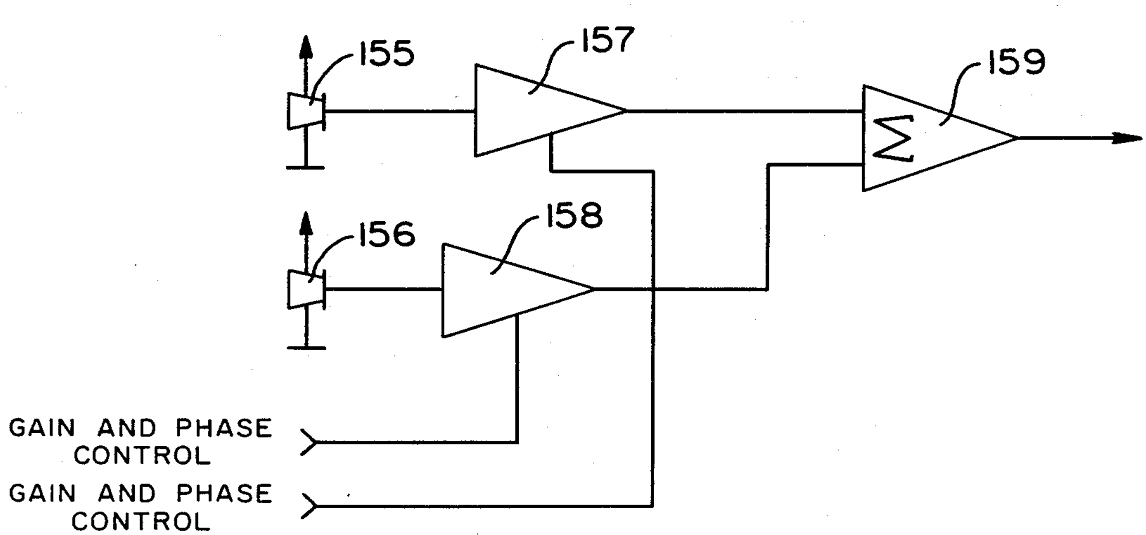

VOLUME
FILTER POSITION SELECT
FBK
P
FWD
R
C
ON/OFF
1,5V
OSC. DIVIDER
VOLTAGE TRIPLER TO ALL CKTS.
UPDATE
LPF
TIMING LOGIC POWER UP RESET
5 BIT UP/DWN. COUNTER
12 BIT A/D
32 x 12 RAM
12 BIT REG.(TEMP.)
2048 x 12 RAM
PROGR. LIMITER
S/H
CAP. SUMMING CKT.
SIGN
12 BIT D/A
7 BIT D/A
V_{REF}
LATCH
CK
DECODER 4 TO 2
SPEECH DETECTOR
BPF
FIXED FILTER
Σ
PROGR. AGC
TRI STATE SW.
11 BIT COUNTER
POWER UP COUNTER CKT.
TRI STATE SW
÷12
SER. PAR. CONV.
EEPROM
D_7
D_{0-6}
D_{8-9}
D_{10-11}
A_{0-4}
A_{5-10}
RESET
CLOCK
V_{CC}
$V_{\phi\ OUT}$
PROG./OP a
PROG./OP b
MIC. ON/OFF
V_{FBK}
NUL
$V_{TEST\ IN}$

FIG. 2



50
40
30
20
10
GAIN (dB)
A
B
C
D
.125
.25
.5
1
2
4
8
FREQUENCY (kHz)

FIG. 3



155
157
159
156
158
GAIN AND PHASE CONTROL
GAIN AND PHASE CONTROL

FIG. 5

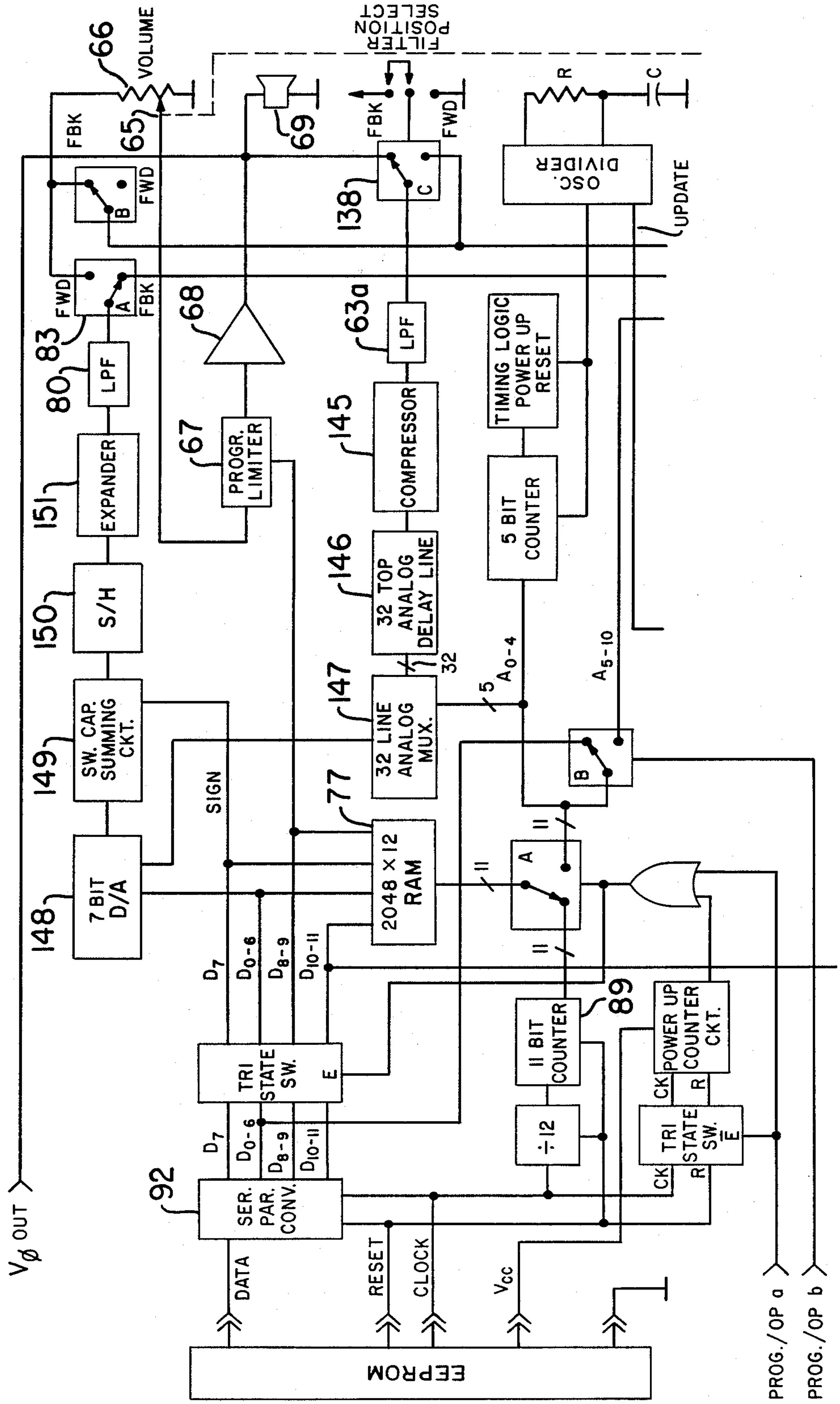

VOLUME
FILTER POSITION SELECT
FBK
FWD
OSC. DIVIDER
UPDATE
LPF
PROGR. LIMITER
COMPRESSOR
TIMING LOGIC POWER UP RESET
EXPANDER
5 BIT COUNTER
32 TOP ANALOG DELAY LINE
S/H
32 LINE ANALOG MUX.
SW. CAP. SUMMING CKT.
SIGN
2048 × 12 RAM
7 BIT D/A
11 BIT COUNTER
POWER UP COUNTER CKT.
TRI STATE SW.
÷ 12
SER. PAR. CONV.
DATA
RESET
CLOCK
EEPROM
PROG./OP a
PROG./OP b

FIG. 4

PROGRAMMABLE DIGITAL HEARING AID SYSTEM

This invention relates to hearing aids, and more particularly to hearing aids that are programmable so as to have suitable characteristics to compensate for the hearing deficiencies of a patient. More specifically, it relates to hearing aids of this character that are capable of automatically adjusting to optimum parameter values as operating conditions such as speech level, room reverberation and background noise change, and also for reducing acoustic feedback.

BACKGROUND OF THE INVENTION

Conventional hearing aids suffer from several shortcomings. It is difficult if not impossible with conventional hearing aids to provide a frequency-gain characteristic that is ideal for each individual user. The acoustic coupling between the hearing aid receiver and the eardrum also introduces changes in the frequency-gain characteristic that is usually deleterious to both speech intelligibility and overall sound quality. For many patients, the optimum frequency-gain characteristic varies as a function of the level of the speech signal reaching the hearing aid. In order to protect patients from uncomfortably or dangerously loud signals, it is also necessary to limit the maximum acoustic power output of the hearing aid in some way. The methods used to limit acoustic power output of hearing aids typically introduce deleterious distortions to the amplified speech signal.

Another common problem is that of acoustic feedback. Even in the best designed hearing aids, not all of the amplified acoustic signal is delivered to the eardrum. A small proportion of the amplified acoustic signal leaks back to the hearing aid microphone forming an acoustic feedback loop. If the gain of the hearing aid is sufficiently high, this acoustic feedback will cause a self-generating oscillation to occur, resulting in an unwanted and highly unpleasant whistling sound. These acoustic oscillations prevent the hearing aid from being used. Methods of acoustic feedback control that are typically used include a tighter acoustic seal between the earmold and the walls of the ear canal so as to reduce acoustic leakage, placing the microphone at some distance from the hearing aid receiver, e.g. on the opposite ear, or simply reducing the gain of the hearing aid. None of these methods provides a satisfactory solution for high-gain hearing aids.

One of the most common complaints of hearing aid users is that background noise is particularly damaging to the understanding of speech. Methods currently used to reduce background noise in hearing aids employ filtering techniques in which the frequency regions containing high noise levels are eliminated.

Another common problem is that of room reverberation produced by acoustic reflections off the walls, ceiling, floor, and other surfaces in a room. A small amount of reverberation is beneficial but too much reverberation will make a room sound hollow or echoic and will interfere with both the quality and the intelligibility of speech.

Systems have been proposed heretofore utilizing computers for testing the hearing of patients and generating programming for a programmable hearing aid, as disclosed in "Computer Application in Audiology and Rehabilitation of the Hearing Impaired" by Harry Levitt, Journal of Communication Disorders 13 (1980), pages 471-481, and in the U.S. Pat. Nos. 4,187,413 to Moser, 4,489,610 to Slavin, and 4,548,082 to Engebretson et al., for example. None of these systems, however, affords a satisfactory solution for the problems of acoustic feedback, background noise, room reverberation and changes in the optimum frequency-gain characteristic resulting from variations in the level of the speech signal reaching the hearing aid.

It is an object of the invention, accordingly, to provide a new and improved hearing aid system that is free of the above-noted deficiencies of the prior art.

Another object of the invention is to provide new and improved hearing aid apparatus of the above character which is capable of automatically adjusting to an optimum set of parameter values as the speech level and type of background noise change.

A further object of the invention is to provide new and improved hearing aid apparatus of the above character in which acoustic feedback is substantially reduced.

Still another object of the invention is to provide new and improved hearing aid apparatus that is capable of effective noise and reverberation suppression and acoustic feedback reduction while maintaining optimum hearing characteristics as the speech and noise levels vary.

SUMMARY OF THE INVENTION

A hearing aid system according to the invention comprises a hearing aid that is programmable so as to have optimum electro-acoustic characteristics for the patient and acoustic environment in which it is used. It also includes instrumentation for measuring relevant audiological characteristics of the patient, as well as techniques and instrumentation for programming the hearing aid to have selected characteristics to compensate for hearing deficiencies determined from the measurements made. Desirably, several sets of optimum hearing aid parameter values, specified in terms of both the amplitude and phase characteristics, are determined for the patient as a function of speech level, type of background noise, and room reverberation, both spectral and temporal characteristics of the noise being taken into account. The selected optimum parameter values are preferably programmed into an electronically erasable, programmable read only memory (EEPROM) which supplies coefficients to programmable filter and amplitude limiting means in the hearing aid so as to cause the hearing aid to adjust automatically to the optimum set of parameter values for the speech level, room reverberation, and type of background noise then obtaining.

In one form, the programmable filter may be a digital equivalent of a tapped delay line in which each delayed sample is multiplied by a weighting coefficient. The sum of the weighted samples generates the desired electroacoustic characteristics. Alternatively, the programmable filter may be a tapped analog delay line in which the sum of the weighted outputs of the taps generates the desired characteristics. An important advantage of the latter type of filter is that the power consumption is low and quasi-digital techniques can be used, i.e., the waveform is sampled at discrete intervals in time without analog-to-digital conversion.

Another form of filter uses a small number of delays in which the delayed output is multiplied by a coeffici-

ent and added to the filter input so as to achieve additional delays, a technique known as recursive filtering.

The invention also provides means for reducing acoustical feedback. In one embodiment, an electrical feedback path in the hearing aid is matched in both amplitude and phase to the acoustic feedback path and the two feedback signals are subtracted so as to cancel each other. In an alternative embodiment, a single filter in the forward path is used with a transmission characteristic equivalent to that of the filter in the forward path plus the electrical feedback path.

Environmental noise control is effected according to the invention by providing means for sensing the relative speech/noise content in the signals from the hearing aid microphone and generating binary words that are supplied to the programmable filter for selecting from a memory a set of delay line tap coefficients that are effective to impart to the filter the appropriate frequency response for the specific environmental noise condition then being detected.

According to the invention, reduction in both noise and reverberation is achieved by the use of two or more microphones. The output of each microphone is processed in both amplitude and phase such that the summed output of the microphones is analogous to the output of a frequency selective directional microphone.

DESCRIPTION OF THE PREFERRED EMBODIMENTS

For a better understanding of the invention, reference is made to the following detailed description of a representative embodiment taken in conjunction with the accompanying drawings, in which:

FIG. 1 illustrates schematically a host controller for use in prescribing a wearable, programmable hearing aid according to the invention;

FIG. 2 illustrates schematically one form of programmable hearing aid according to the invention which utilizes a digital delay line filter;

FIG. 3 shows a set of optimum frequency-gain characteristics appropriate for different speech levels for a typical hearing impaired subject;

FIG. 4 is a partial schematic diagram of the hearing aid shown in FIG. 2, modified to utilize an analog delay line filter; and

FIG. 5 shows schematically how multiple microphones can be used according to the invention to reduce both noise and reverberation.

A hearing aid system according to the invention comprises generally a wearable, programmable hearing aid in which all operations are controlled by data stored in an erasable electrical programmable read only memory (EEPROM), and a host controller providing electrical signals and test sounds as necessary for measuring the residual hearing of a subject, establishing optimal hearing aid parameters for the subject (including the phase relationship between input and output) and generating control signals as necessary to program the EEPROM module to perform the desired operations in the hearing aid.

THE HOST CONTROLLER

Referring first to FIG. 1, the host controller 20 is shown as comprising an audiometric signal generator 21 including a signal generator 22, the frequency of which is controllable by a digital-to-analog (D/A) converter 23 controlled by a computer 24 through a latch 25. The output level of the signal generator is adapted to be

adjusted by an attenuator 26 also controlled by the computer 24, through a latch 27. The output of the attenuator 26 is supplied to one terminal 28 of a switch 29, from which it can be transmitted through an amplifier 119 and a switch 120 to a terminal connector 121.

The host controller 20 also includes a phase measurement circuit comprising a digital phase shifter 30 which is adapted to receive the output voltage of the hearing aid through a connector 31 and a programmable gain amplifier 32, the gain of which is controllable by the computer 24 through the latch 33.

In adjusting a hearing aid for reduced feedback as described below, the input voltage developed by the hearing aid microphone in response to acoustic feedback alone when a test signal is supplied to the hearing aid is summed with an electrical feedback voltage supplied to the terminal 144 by the phase shifter 30. The summed acoustic and electrical feedback voltages from the hearing aid are supplied from the terminal 34 through the engaged movable and fixed contacts 35 and 36 of a switch 37, a programmable gain amplifier 38, a rectifier 39 and an analog to digital (A/D) converter 40 to the computer 24. In effecting the adjustment, the gain of the amplifier 32 and the phase shifter 30 are adjusted until a null output from the A/D converter 40 is read by the computer 24, indicating cancellation of the electrical and acoustic signals. The settings of the phase shifter 30 and the amplifier are then used to program an EEPROM in the hearing aid so as to cancel acoustic feedback.

The programmable gain amplifier 38 is controlled by the computer 24 through a latch 41, and the digital phase shifter 30 is also controlled by the computer 24 through the latches 42 and 43.

Additional components of the host controller 20 are programming logic comprising the latches 44 and 45 and a conventional tri-state buffer 46 which are activated in response to the computer 24 to provide the necessary signals to program the EEPROM and control the hearing aid during testing and programming, as described in greater detail below.

A conventional tape player 47 is also provided for generating various sound combinations for testing of a patient's hearing. The tape player 47 is controlled by the computer 24 through a latch 48 and its output level is adapted to be controlled by an attenuator 49, also controlled by the computer 24, through a latch 50. The output of the attenuator 49 is supplied to the movable contact 51 of a switch 52 which is controllable to supply the output either to a connector 53 to a sound field system, or to a fixed contact 54 on the switch 29 for supply to the hearing aid, as described in greater detail below. The switches 29 and 52 and the tri-state buffer 46 are adapted to be controlled by the computer 24 through the latch 44.

The respective latches 27, 25, 42, 43, 33, 40, 41, 44, 45, 48 and 50 are adapted to be controlled by the computer 24 through an address decoder 55 and Read/Write logic 56.

THE HEARING AID

Referring now to FIG. 2, a wearable hearing aid according to the invention comprises a microphone 57, the output of which is fed through a programmable automatic gain control (AGC) circuit 58 and a switch 59 to one terminal of summing amplifier 60. In normal operation, amplifier 60 supplies the signal through the conductors 61 and 62 and a filter 63 to a programmable

filter **64** which is adapted to be programmed in the manner described below to produce optimum hearing aid characteristics for the patient based on the measurements made of the patient's residual hearing. The output of the programmable filter **64** is fed from the movable contact **65** of the volume control **66**, through a programmable limiter **67**, and an amplifier **68** to the hearing aid receiver **69**.

The programmable filter **64** comprises essentially a digital tapped delay line including a 12 bit A/D converter **70**, which supplies outputs to a 32×12 random access memory (RAM) **71**, a 12 bit temporary register **72**, and a 12 bit D/A converter **73**. The signal supplied to the RAM **71** is sampled by a 5 bit up-down counter **74** controlled by timing logic **75** connected to receive clock signals from an oscillator divider **76** oscillating at a frequency at least twice the audio signal band width.

The characteristics of the programmable filter **64** are determined by coefficients stored in a random access memory (RAM) **77** which are selected and fed to a 7 bit D/A converter **78**, the output of which is supplied to the D/A converter **73** for multiplication of the sampled data by a selected coefficient from the RAM **77**. The output of the D/A converter **73** is summed in a conventional charge transfer summing circuit **78*a*** and the sum signal is supplied through a sample and hold circuit **79** to a conventional anti-imaging filter **80**. The output of the filter **80** is fed through the movable contact **81** of a switch **83** and a fixed contact **82** thereof to the volume control **66** and eventually to the receiver **69**.

The coefficients stored in the RAM **77** when the hearing aid is in operation are provided essentially by an EEPROM **84** previously programmed by the host controller **20** (FIG. 1) as described in greater detail hereinafter. On power-up, the filter coefficients and limit parameters are transferred from the EEPROM **84** to the RAM **77** as follows: Tri-state switches **85** and **86** are enabled and power is supplied to the EEPROM **84** from a power-up control circuit **87**. A switch **88** is now activated to connect an 11 bit counter **89** to the RAM **77** through the switch movable contact **90**. The power-up control circuit **87** acting through the tri-state switch **85** supplies reset pulses to a divide-by-12 counter **91**, to the 11 bit counter **89** and to a series parallel converter **92**.

The power-up control circuit **87** also supplies clock pulses to the series parallel converter **92** and to the divide-by-12 counter **91** such that after every twelfth clock pulse, data is transferred from the EEPROM **84** through the series parallel converter **92** and the tri-state switch **86** to one of the memory locations in the RAM **77** as determined by the 11 bit counter **89**. These steps are repeated as many times as are required to transfer all of the data stored in the EEPROM **84** to the RAM **77**. When that occurs, the tri-state switch **86** is disabled and supplies a signal to the switch **88** to connect the movable contact **90** thereof to the fixed contact **93**, which is connected to the movable contact **94** of a switch **95** and to the 5 bit counter **74** and the RAM **71**. The hearing aid is then in its normal operating mode.

Automatic adjustment of the hearing aid to take into account environmental conditions such as changes in the speech level and type of background noise obtaining at any time is effected by environmental control means comprising a speech detector **96**, four band pass filters **97, 98, 99** and **100**, and a level detector including four differently prebiased comparators **101, 102, 103** and **104**. Typical bandwidths for the filters **97, 98, 99** and **100** might be 100–750 Hz, 750–1500 Kz, 1500–3000 Hz, and 3000 to the upper frequency limit, respectively. The speech detector **96** is conventional and may include a level detector followed by a short term averaging device. For steady state noise, the output of the level detector will be relatively constant, indicating that noise only is present. Whenever fluctuations in level are within a prescribed bandwidth typical of speech, the short term average increases, indicating that speech is present. The speech detector **96** is clocked periodically by signals from the oscillator divider **76** to provide either a zero output indicative of noise or a one output indicative of speech periodically to each of a plurality of sample and hold circuits **105, 106, 107** and **108**. The outputs of the band pass filters **97, 98, 99** and **100** are rectified in the rectifiers **109, 110, 111** and **112**, respectively and fed to the sample and hold circuits **105–108**, respectively, the outputs of which are fed to the comparators **113, 114, 115** and **116**, respectively. The outputs of the rectifiers **109–112**, respectively, are also supplied directly as inputs to the comparators **113–116**, respectively.

So long as there is a one output from the speech detector **96** indicating the presence of speech in the input, the instantaneous outputs of each of the band pass filters **97–100** are compared with previous values held in the sample and hold circuits **105–108**, causing the comparators **113–116**, respectively, to generate a binary coefficient (0,1) indicating whether or not the speech level in the associated band pass filter exceeds the noise level. At the same time, a level detector **117** responsive to the respective outputs of the comparators **101–104** generates a two bit coefficient indicating the average signal level.

The outputs of the comparators **113–116**, inclusive, and of the input level detector **117** are fed to latching means **118**, which provides a six bit output that is adapted to be transmitted through the switches **95** and **88** to the RAM **77**. Whenever the output of the speech detector **96** indicates that speech is present, the binary outputs of the four filter level comparators **113–116**, inclusive, and that of the average level detector **117** are transmitted to the RAM **77** through the control switches **95** and **88**. These binary words are updated at regular intervals at a clock rate determined by the oscillator divider **76**, as stated. Each of the sixty-four possible combinations of the 6 bit binary words identifies a different frequency response for the programmable filter, and a corresponding set of coefficients stored in the RAM **77** is selected, thereby automatically adjusting the hearing aid to the optimum set of parameter values as the speech level and type of background noise change.

In prescribing hearing aids with the use of the host controller, the subject is seated in a quiet room with the hearing aid inserted in his ear. The hearing aid is connected to the host controller by an electrical cable (not shown), thereby placing it directly under the control of the host controller. The prescriptive procedure usually consists of five stages:

(1) measurement of the subject's residual hearing;

(2) derivation of an appropriate set of electroacoustic characteristics of the hearing aid from such measurements, and programming the hearing aid accordingly;

(3) measuring acoustic feedback in the hearing aid;

(4) re-programming the hearing aid to minimize acoustic feedback; and

(5) paired comparison testing of possible alternative settings of the hearing aid to determine the optimal hearing aid settings for the subject.

The subject's residual hearing is measured using signals generated by the audiometric section **21** in the host controller **20** (FIG. **1**). These signals are delivered to the hearing aid via the switch **29**, an amplifier **119**, a switch **120**, a connector **121**, a connector **122** in the hearing aid, and the conductor **62**. The signals are processed by the digital programming filter and associated circuitry in the hearing aid and are delivered to the subject's ear using the hearing aid receiver and associated coupling that the subject will actually wear after the hearing aid is prescribed. This procedure eliminates the need for any corrections in going from headphone to sound field measurements. The measurements are usually obtained with narrow band stimuli (tones, warble tones, or narrow band noise) and include threshold of hearing, various loudness levels (comfortable, loud) and loudness discomfort level.

The measurements obtained on the patient's residual hearing are used in deriving the electroacoustic characteristics of the hearing aid. The measurements of loudness discomfort level are used to program the limiter **67** so that sounds amplified by the hearing aid never exceed the patient's loudness discomfort level. The measurements of auditory threshold, most comfortable loudness level, and loudness discomfort level are used to determine the frequency gain characteristics of the hearing aid.

FIG. **3** shows four frequency-gain characteristics for a typical patient. Curve A is used when the speech signal reaching the hearing aid is relatively low, as would occur when somebody speaks in a very soft voice. Under these conditions the hearing aid provides a large amount of gain, particularly in the high frequencies. This is done to ensure that the speech spectrum is placed above the patient's threshold of hearing at all frequencies.

Curve B is used when the incoming speech signal is at the low end of the comfortable loudness range for a normal hearing person. The amplification provided places the speech spectrum at the bottom of the patient's most comfortable loudness range at all frequencies.

Curve C is used when the level of the incoming speech is moderately loud for a normal hearing person. The amplification provided places the speech spectrum at the top of the patient's most comfortable loudness range for all but the lowest frequencies. Less gain is provided at the low frequencies to reduce upward spread of masking effects; e.g., weak high-frequency sounds being masked by intense low frequency sounds.

Curve D is used when the signals reaching the hearing aid are very loud for a normal hearing person. Under these conditions the hearing aid provides relatively little gain with a significant roll off in the low frequency region in order to substantially reduce upward spread of masking effects.

A set of coefficients is derived for each of these frequency gain characteristics. These coefficients are derived using procedures that are well known in the field of digital signal filtering and are used to program filter **64** so that the hearing aid produces the required frequency-gain characteristic. The filter coefficients are stored in the RAM **77**.

The level of the incoming speech signal is determined by the level detectors **101**, **102**, **103** and **104**. The decoder **117** generates a binary word depending on the outputs of these level detectors. This binary word is transmitted to the RAM **77**, in order to select the appropriate set of filter coefficients. If the signal reaching the hearing aid consists of speech plus noise, as determined by the speech detector **96**, then alternative frequency-gain characteristics are used. These frequency-gain characteristics are derived by first determining the incoming signal level, as described above, selecting an appropriate frequency-gain characteristic and then reducing the gain in those frequency regions where the background noise level exceeds the speech level. This is determined by comparing the output levels of the band-pass filters **97**, **98**, **99**, and **100** when speech is present to the corresponding levels measured when noise only is present. The latter information is stored in the sample and hold units **105**, **106**, **107** and **108**. The comparisons are done by means of the comparators **113**, **114**, **115**, and **116**. The outputs of these comparators in combination with the outputs of the level decoder, **117**, generate a 6-bit word that selects the appropriate set of filter coefficients in the RAM **77**.

For patient hearing parameter selection and programming, the hearing aid is interfaced with the host controller as described above, and the EEPROM **84** (FIG. **2**) is plugged into a programming slot **124** in the host controller **20**. A conductor in the line **125** (FIG. **1**) is set to logic **1** by operation of the computer **24** which applies a logic **1** signal to the line **126** (FIG. **2**), resulting in the tri-state switch **85** being disabled and the switch **88** being activated by an OR gate **127** to move the switch movable contact **90** to connect the counter **89** to the RAM **77**. The host controller **20** generates reset pulses which are transmitted over a conductor in the line **128** (FIG. **1**) through the connector **129** and the line **130** in the hearing aid (FIG. **2**) to reset the counters **91** and **89** and the series-to-parallel converter **92**.

Clock signals are fed from the host controller **20** over another conductor in the line **128** through a connector **131** and a conductor **132** in the hearing aid to the serial-to-parallel converter **92** and are divided by 12 in the counter **91**, the output of which is fed through the switch **88** to the RAM **77**. Synchronously with the clock signals on the line **132**, data are fed from a conductor in the line **128** in the host controller **20** and through a connector **133** and a conductor **134** in the hearing aid to the serial-to-parallel converter **92**. After every twelfth clock signal, data is transferred from the serial-to-parallel converter **92** through the tri-state switch **86** to one of the memory locations in the RAM **77** determined by the counter **89**. This step may be repeated as many times as required to store essential data in the RAM **77** within the storage limits of the latter. The line **126** is then set to logic **0** by operation of the computer **24** through a conductor in the line **125**, thus disabling the tristate switch **86** and actuating the switch **88** to cause the movable contact **90** thereof to move into engagement with the fixed contact **93** resulting in the connection of the RAM **77** to the counter **74** and to the RAM **71**. The hearing aid is now ready for patient hearing parameter selection and programming.

The selection of the desired hearing aid parameters is accomplished by actuating the computer **24** to set the line **135** in the hearing aid (FIG. **2**) to logic **1** through a conductor in the cable **128** and a connector **136** on the hearing aid. This activates the switch **95** and supplies data in the form of 6 bit words from the series-to-parallel converter **92** over the line **137** through the movable

contact **94** of the switch **95** and the movable contact **90** of the switch **88** to the RAM **77**. The series-to-parallel converter **92** is reset by a signal from the computer **24** (FIG. **1**) transmitted through a conductor in the line **128** and the connector **129** and conductor **130** in the hearing aid, and a 6 bit word is fed into the converter **92** and placed on seven of the address lines of the RAM **77**, selecting one of sixty-four possible sets of coefficients for use in the programmable filter in the hearing aid. The selection proceeds in the like fashion throughout the process of frequency shaping of the filter.

Feedback cancellation is achieved in the hearing aid by first measuring the acoustic feed back in situ and then creating an electronic feedback path with identical amplitude and phase characteristics. The outputs of the two feedback paths are then subtracted, thereby canceling any feedback signals that might occur.

Acoustic feedback is measured with the hearing aid in the ear as it would normally be worn. An electrical test signal is applied to the terminal **122** (FIG. **2**) from the host controller **20**. A portion of this amplified acoustic signal will leak through the ear mold and reach the microphone **57**, which will then convert the signal back to electrical form and return it to the hearing aid amplifier. Feedback will occur if the total gain in the loop, i.e., from the input to the filter **63** through the filter and amplifier of the hearing aid to the output transducer **69** and from the microphone **57** back to the input to the filter **63**, exceeds unity.

For the purpose of this measurement, the feedback loop is broken between **140** and **122**. Terminal **140** of the hearing aid is then connected to terminal **34** of the host controller, terminal **142** of the hearing aid is connected to terminal **31** of the host controller, and terminal **143** of the hearing aid is connected to terminal **144** of the host controller. The programmable phase shifter **30** and programmable amplifier **32** are then adjusted by the computer **24** so as to minimize the sum of the acoustic and electrical feedback signals of the output of summing amplifier **60**.

From the settings obtained with the phase shifter **30** and amplifier **32** at cancellation (at the "null"), the host controller calculates a set of coefficients for a programmable filter to be inserted in the electrical feedback path between terminals **142** and **143** so as to cancel the acoustic feedback.

If desired, the additional programmable filter in the feedback path of the hearing aid can be eliminated by calculating the effective gain of both the forward and feedback paths and modifying the programmable filter **64** so as to include this additional gain. This implementation necessarily requires an adjustment to both the amplitude and phase characteristics of the filter **64**. Alternatively, filter **64** can be placed in the feedback path with appropriate changes in the coefficients.

The setting of the hearing aid is then checked using a paired comparison technique to determine if optimum frequency response characteristics for the patient have been obtained. This involves the provision in the host controller of one or more additional programmable filters adjusted to have frequency responses that differ systematically from that prescribed using the above procedure. In the practice of this technique, the patient compares the prescribed hearing aid against an alternative hearing aid by replacing the prescribed filter and limiter with components having alternative electro-acoustic characteristics by actuation of a select switch **123** in the host controller **20**. The patient listens to

speech through the hearing aid, switching back and forth two sets of electroacoustic characteristics at will by means of the switches **123A** and **123B**, choosing the characteristic which is more intelligible or preferable in some way. Paired comparisons made in this manner may be repeated with systematic deviations from the prescribed frequency response in order to determine whether another frequency response would be more intelligible or preferable. If a better response is found, the paired comparison procedure may be repeated iteratively until the optimum set of electroacoustic characteristics is found. Adaptive strategies for finding the optimum electro-acoustic characteristics are described in Levitt et al., Journal of Rehabilitation Research and Development, Vol. 23, No. 1, 1986, pages 79–87.

In normal operation, the EEPROM **84**, with the optimum coefficients determined as described above stored therein, is plugged into the hearing aid, and when power is turned on the coefficients are transferred from the EEPROM **84** into the RAM **77**. From then on the EEPROM is inactive in order to save power. The operating sequence is as follows: On power-up the tri-state switches **85** and **86** are enabled, power is supplied to the EEPROM **84** from the power-up control circuit **87**, and the switch **88** is activated to connect the counter **89** to the RAM **77**. The power-up circuit control **87** supplies reset pulses to the counters **89** and **91** and to the series-to-parallel converter **92**, and clock pulses to the counter **91**, the converter **92** and the EEPROM **84**. After every twelfth clock pulse, data is transferred from the series-to-parallel converter **92** through the tri-state switch **86** to one of the memory locations in the RAM **77** determined by the counter **89**. This step is repeated until all the data stored in the EEPROM **84** have been transferred to the RAM **77**. The tri-state switch **86** is then disabled and the switch **88** is connected to the switch **37**, the counter **74** and the RAM **71**.

The hearing aid is now in its normal operating mode and speech detected by the microphone **57** is amplified in the programmable automatic gain control circuit **58** and transmitted through the amplifier **60**, the filter **63** and the low pass filter ***63a*** into the programmable filter circuitry.

A so-called "bucket brigade" analog delay line may be used as a programmable filter instead of the digital delay line described above, as shown in FIG. **4**. Thus, the audio signal may be fed from the filter ***63a*** through a compressor **145** to a delay line **146** having, say, thirty-two taps therealong. The delay line **146** is clocked by a 50% duty cycle signal at a frequency equal to at least twice the audio signal bandwidth. During the logic "1" period of the clock, the audio signal is supplied into the next stage of the analog delay line. On the falling edge of the clock signal, the analog signal is held in the next stage. During the logic "0" period of the clock, an analog multiplexer **147** selects one of the thirty-two taps of the delay line and feeds the signal therefrom into the input of a 7 bit D/A converter **148**. The signal is divided down in a resistance ladder in the converter **148** as determined by a 7 bit word from the RAM **77**, the address of which is specified by the counter **89**.

The divided down voltage is then given a plus or minus sign as determined by a bit from the series-parallel converter **92** and is fed to a switched capacitor summing circuit **149**. The analog multiplexer **147** proceeds to the next tap and repeats the process just described for a total of thirty-two times. The thirty-two voltages thus derived are summed in the summing circuit **149** and

transferred to a sample and hold circuit **150**, the output of which is expanded in an expander **151** and supplied through the filter **80** to the hearing aid receiver **69**.

An analog tapped delay line in which the delays are achieved acoustically might also be employed in a programmable hearing aid filter according to the invention, instead of the digital and analog delay lines described above. This might be accomplished by feeding the hearing aid electrical signals to a transducer to generate sound to travel along a tube of appropriate length and having taps in the form of miniature microphones disposed along the tube. The signals at the taps would be multiplied by selected predetermined coefficients and added to produce a resultant characteristic much in the same manner as described above.

By using an array of two or more microphones on the body (e.g., along the frame of a pair of spectacles), the weighting coefficients can be chosen in the manner described above such that the weighted summed output of the microphones, with an appropriate phase shift, is equivalent to the output of a frequency selective directional microphone. This will reduce the effects of both noise and room reverberation. A typical configuration is shown in FIG. **5** as comprising the microphones **155** and **156** supplying inputs to amplifiers **157** and **158**, the gain and phase characteristics of each of which is programmable from the RAM **77** in the hearing aid. The outputs of the programmable gain amplifiers **157** and **158** are summed in a summing amplifier **159**, the output of which would be supplied to the programmable AGC device **58**.

Automatic adjustment of the frequency response of the hearing aid as a function of speech level may also be effected by placing the programmable filter in a feedback loop in series with a programmable compression amplifier. Since the degree of feedback depends on signal level, the overall frequency response also depends on signal level.

The components shown in FIGS. **2**, **4** and **5** may be incorporated in the hearing aid or part may be contained in a pocket size case to be carried in the clothing of the person wearing the hearing aid. In the latter case, the components contained in the case may be coupled to the components in the hearing aid by a conventional FM transmission link, for example.

The invention thus provides a novel and highly effective hearing aid system for use by hearing impaired patients. By utilizing a digital or analog delay line as a programmable filter in a hearing aid, it is possible to establish optimal hearing aid parameters for the patient. Moreover, by virtue of the novel means employed for effecting automatic adjustment of the programmable filter to optimum parameter values as the speech level, room reverberation and type of background noise change and for reducing acoustic feedback, a superior hearing aid system of optimum characteristics can be prescribed for hearing deficient patients.

The several embodiments described above and depicted in the drawings are intended to be only illustrative, and modifications in form and detail are possible within the scope of the following claims. For example, more than four bandwidth filters and different frequency bands might be employed. Also, the number of taps on the digital filter may be greater than 32 as in the specific embodiment described above and the A/D converter may, of course have more than 12 bit resolution.

We claim:

1. A hearing aid comprising at least one input microphone, an output receiver, and a signal transmission channel interposed between said microphone and said receiver in which the improvement comprises a programmable delay line filter and programmable signal limiter means interposed between the input and output of said transmission channel in a feedback path for said transmission channel, said filter being programmed to impart to the hearing aid at least one response characteristic effective to compensate for impaired hearing of the wearer of the aid.

2. A hearing aid as in claim **1** in which said filter comprises a delay line having a multiplicity of taps, first memory means for storing filter-response related coefficients therein, and means jointly responsive to said delay line taps and to coefficients stored in said memory means for providing an output having at least one response characteristic effective to compensate for impaired hearing of the wearer of the aid.

3. A hearing aid as in claim **2** in which said filter comprises a digital delay line and means for converting analog signals in said channel to digital signals for application to said delay line.

4. A hearing aid as in claim **2** in which said filter comprises an analog delay line having a multiplicity of taps, and means for periodically sampling analog signals in said channel and supplying them to said delay line.

5. A hearing aid comprising at least one input microphone, an output receiver, and a transmission channel interposed between said microphone and receiver, means controllable to impart different response characteristics to said hearing aid, and controlling means responsive to the level of speech signals in said transmission channel in excess of the level of noise signals in said channel for automatically controlling said controllable means to impart a selected one of said different response characteristics to said hearing aid.

6. A hearing aid as described in claim **5** in which said controlling means comprises speech detector means for determining when speech signals are in said transmission channel, a plurality of bandpass filter means for determining the noise frequency spectrum in said transmission channel, a plurality of comparator means each response to said speech detector and to respective bandpass filter means for indicating whether the speech level in each said bandpass filter exceeds the noise level therein and for actuating said controlling means to impart to the hearing aid a response characteristic effective to compensate for impaired hearing of the wearer of the aid at the noise levels obtaining in said channel.

7. A hearing aid as described in claim **6** in which said controlling means also comprises means for rectifying a portion of the signals in said transmission channel, a plurality of differently pre-biased comparators connected to receive the output of said rectifying means, and means responsive to the outputs of said comparators for generating a signal representative of the average level of the signals in said transmission channel for controlling said controllable means to impart to the hearing aid a response characteristic effective to compensate for impaired hearing of the wearer of the aid at the signal and noise levels obtaining in said channel.

8. A hearing aid as described in claim **7** in which said controllable means comprises a delay line having a multiplicity of taps, memory means for storing predetermined filter response-related coefficients therein, and means responsive to said controlling means for combining said delay line taps and coefficients to constitute a

delay line filter of predetermined response characteristic in said transmission channel.

9. A hearing aid comprising at least one input microphone, an output receiver, a signal transmission channel interposed between said microphone and said receiver, and a programmable delay line filter interposed in a forward path between the input and output of said transmission channel, said programmable filter being programmed to effect substantial reduction of acoustic feedback from said receiver to said microphone.

10. A hearing aid as described in claim 1 in which said programmable delay line filter is programmed so as to effect substantial reduction of acoustic feedback from said receiver to said microphone.

11. A hearing aid comprising at least one input microphone, an output receiver, a signal transmission channel interposed between said microphone and said receiver, and a programmable delay line filter interposed in a forward path between the input and output of said transmission channel said filter comprising a digital delay line having a multiplicity of taps, means for converting analog signals in said channel to digital signals for application to said delay line, and first memory means for storing filter-response related coefficients therein, said digital delay line comprising second memory means arranged to receive digital signals from said analog-to-digital converting means, register means connected to receive digital signals from said second memory means, first digital-to-analog converter means connected to receive digital signals from said register means, timing means controlling the transfer of digital signals from said second memory means through said register means to said first digital-to-analog converter means, second digital-to-analog converter means connected to receive filter response related coefficients from said memory means in timed relation to the transfer of digital signals to said first digital-to-analog converter means and to supply analog signals therefrom to said first digital-to-analog converter means for combination therein with digital signals transferred thereto from said second memory means, signal summing means connected to receive the output of said first digital converter means, and sample and hold means connected to receive the output of said signal summing means and for providing an output having at least one response characteristic effective to compensate for impaired hearing of the wearer of the aid.

12. A hearing aid comprising at least one input microphone, an output receiver, a signal transmission channel interposed between said microphone and said receiver, and a programmable delay line filter interposed in a forward path between the input and output of aid transmission channel, said filter comprising an analog delay line having a multiplicity of taps, means for periodically sampling analog signals in said channel and sampling analog signals in said channel and supplying them to said delay line, and first memory means for storing filter-response related coeffients therein, digital-to-analog converter means connected to receive filter-response related coefficients from said first memory means for conversion to analog values, multiplexer means connected to supply signal samples from said sampling means to said digital converter means for combination with said respective analog values, summing means for summing said analog values and signal samples, and sample and hold means connected to receive the output of said summing means and for providing an output having at least one response characteristic effective to compensate for impaired hearing of the wearer of the aid.

13. A method of reducing acoustic feedback in a sound system comprising a microphone, a transducer and a signal transmission channel interposed between said microphone and transducer, comprising the steps of

determining the effect on the amplitude and phase of a signal in said transmission channel as a function of frequency of acoustic feedback between said transducer and microphone, and

inserting between the input and output of said transmission channel an electrical feedback path having a filter therein programmed to equalize and reduce the effect of said acoustic feedback both in amplitude and phase on a signal in said transmission channel.

14. A method of reducing acoustic feedback in a hearing aid comprising a microphone, a receiver fitted in an ear of a wearer of the aid, and a signal transmission channel interposed between said microphone and transducer, comprising the steps of

determining the effect on the amplitude and phase of a signal in said transmission channel as a function of frequency of acoustic feedback between said receiver and microphone, and

inserting between the input and output of said transmission channel a programmable filter programmed to equalize and reduce the effect of said acoustic feedback both in amplitude and phase on a signal in said transmission channel.

15. A method of reducing feedback in a hearing aid as described in claim 14 in which said programmable filter is inserted in a forward path through said transmission channel.

16. A method of reducing feedback in a hearing aid as described in claim 14 in which said programmable filter is inserted in an electrical feedback loop for said transmission channel.

17. A hearing aid comprising at least two input microphone channels, means for adjusting the amplitude and phase characteristics of each of said microphone channels, means for summing the outputs of said microphone channels, an output receiver, a signal transmission channel connected to receive the output of said summing means and to provide an output to said receiver, and a programmable filter interposed between said summing means and the output of said transmission channel, said filter being programmed to impart to the hearing aid at least one response characteristic effective to compensate for impaired hearing of the wearer of the aid and to reduce the effects of both noise and reverberation.

18. A hearing aid comprising at least one input microphone, an output receiver, and a signal transmission channel interposed between said microphone and said receiver in which the improvement comprises a signal-level dependent amplifier interposed between the input and output of said transmission channel in a feedback path for said transmission channel, said amplifier being programmed to impart to the hearing aid at least one response characteristic effective to compensate for impaired hearing of the wearer of the aid.

19. A hearing aid comprising at least one input microphone, an output receiver, a signal transmission channel interposed between said microphone and said receiver, and a programmable delay line filter interposed in a feedback path between the input and output of said transmission channel, said programmable filter being programmed to effect substantial reduction of acoustic feedback from said receiver to said microphone.

* * * * *

UNITED STATES PATENT AND TRADEMARK OFFICE

CERTIFICATE OF CORRECTION

PATENT NO. : 4,731,850

DATED : March 15, 1988

INVENTOR(S) : LEVITT ET AL.

It is certified that error appears in the above-identified patent and that said Letters Patent is hereby corrected as shown below:

Col. 5, line 68, "1500 Kz" should read --1500 Hz--; Col. 12, line 44, "response" should read --responsive--; Col. 13, line 52, "aid" should read --said--; Col. 13, line 58, "coeffients" should read --coefficients--.

Signed and Sealed this

Third Day of October, 1989

Attest:

DONALD J. QUIGG

Attesting Officer

Commissioner of Patents and Trademarks

EXHIBIT B

United States Patent [19]

Levitt et al.

[11] **Patent Number:** **4,879,749**

[45] **Date of Patent:** **Nov. 7, 1989**

[54] **HOST CONTROLLER FOR PROGRAMMABLE DIGITAL HEARING AID SYSTEM**

[75] Inventors: **Harry Levitt**, Livingston, N.J.; **Richard S. Dugot**, New York, N.Y.; **Kenneth W. Kopper**, River Edge, N.J.

[73] Assignee: **Audimax, Inc.**, Hackensack, N.J.

[21] Appl. No.: **155,374**

[22] Filed: **Feb. 12, 1988**

Related U.S. Application Data

[62] Division of Ser. No. 879,214, Jun. 26, 1986, Pat. No. 4,731,850.

[51] **Int. Cl.**[4] **H04R 29/00;** H04R 25/00

[52] **U.S. Cl.** **381/68.4;** 73/585; 381/60; 381/68; 381/93

[58] **Field of Search** .... 381/68.4, 68, 60, 83, 381/93; 73/585; 324/73 R, 73 AT, 73 PC

[56] **References Cited**

U.S. PATENT DOCUMENTS

2,263,233	11/1941	Bürck	381/83
2,723,316	11/1955	Goodell et al.	381/83
3,105,877	10/1963	Miller et al.	381/83
3,257,510	6/1966	Burkhard	381/93
3,429,999	2/1969	Stover	381/93
3,651,266	3/1972	Queen	381/83
4,099,035	7/1978	Yanick	381/94
4,109,116	8/1978	Victoreen	381/93
4,191,864	3/1980	Sopher	381/60
4,378,468	3/1983	Braun	381/93
4,449,237	5/1984	Stepp et al.	381/93
4,453,039	6/1984	Ferrieu	381/93
4,471,171	9/1984	Köpke	381/60
4,747,144	5/1988	Admiraal et al.	381/83

FOREIGN PATENT DOCUMENTS

209894	1/1987	European Pat. Off.	381/93
3624764	1/1988	Fed. Rep. of Germany	381/93
1305359	1/1973	United Kingdom	381/93

OTHER PUBLICATIONS

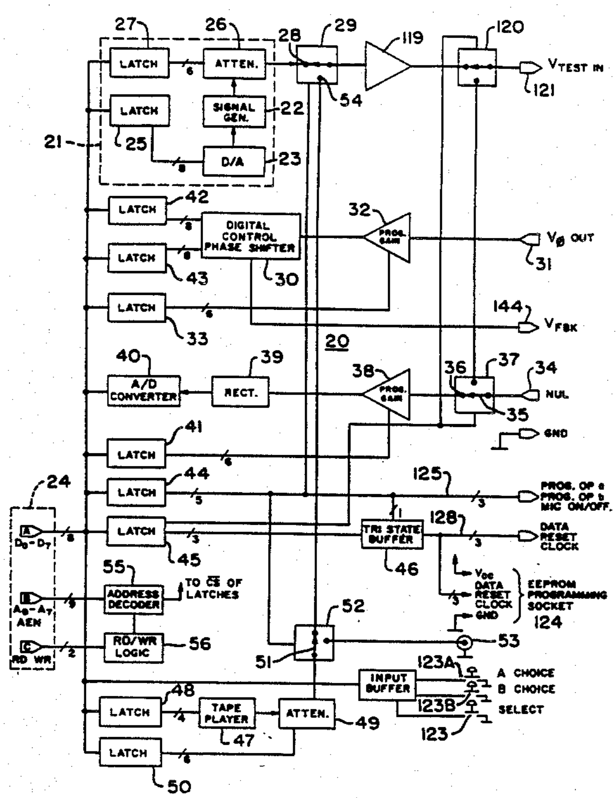
Hearing Instruments, Feb. 1977, Frye, George, J., "Computerized Hearing Aid Testing", pp. 14 and 29.

Primary Examiner—Jin F. Ng
Assistant Examiner—Danita R. Byrd
Attorney, Agent, or Firm—Brumbaugh, Graves, Donohue & Raymond

[57] **ABSTRACT**

A host controller for producing data from a computer for a programmable filter of a hearing aid to cancel feedback in which phase shift and gain control means as adjusted by the computer to generate a feedback cancellation voltage which is supplied to the hearing aid for summation of the feedback and feedback cancellation voltages by the hearing aid and in which the summed feedback and feedback cancellation voltages are returned to the host controller for further adjustment of said phase shift and gain control means until feedback has been cancelled. The host controller then transmits to the programmable filter the phase shift and gain control data necessary to cancel feedback.

6 Claims, 4 Drawing Sheets



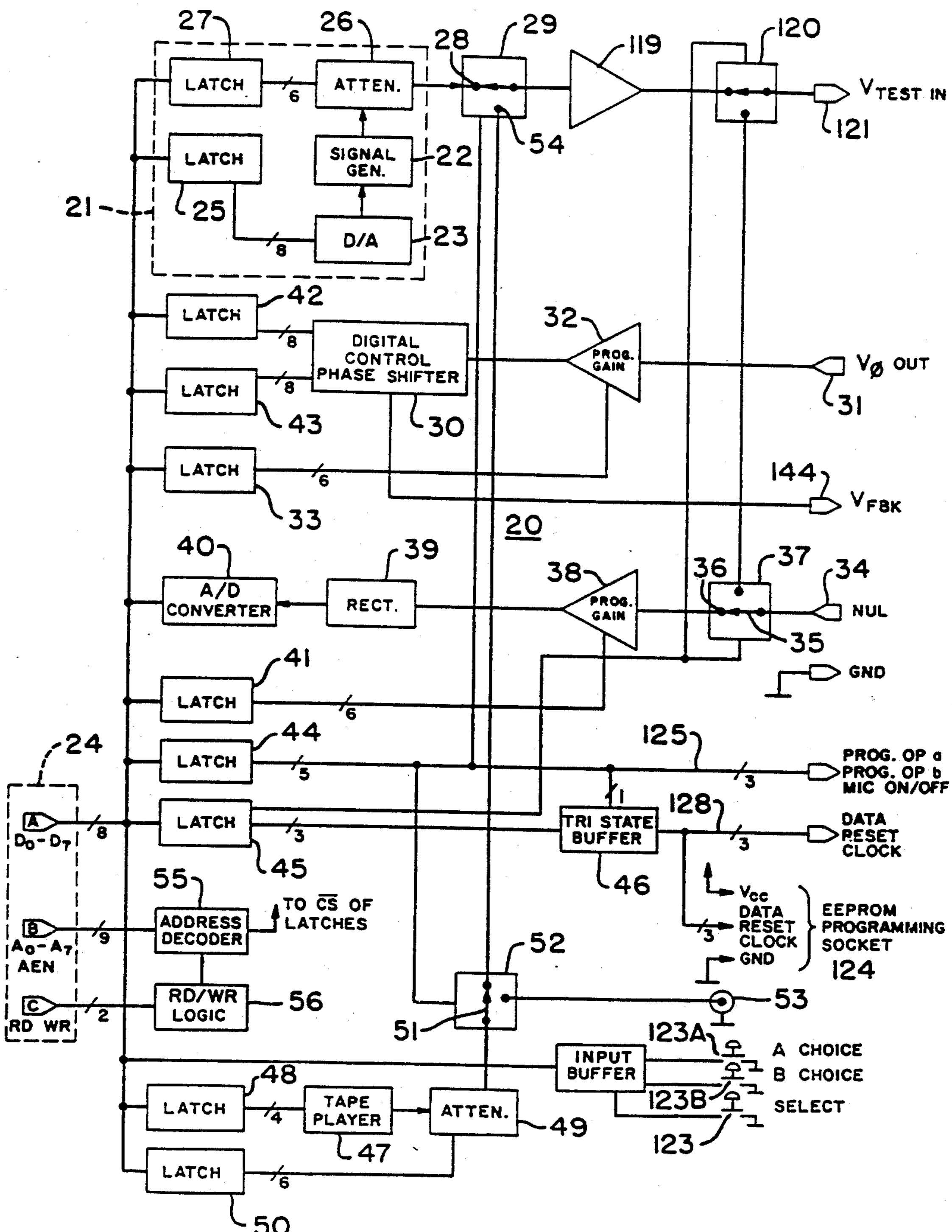

LATCH
ATTEN.
LATCH
SIGNAL GEN.
D/A
LATCH
DIGITAL CONTROL PHASE SHIFTER
LATCH
LATCH
PROG. GAIN
V TEST IN
Vø OUT
V FBK
A/D CONVERTER
RECT.
PROG. GAIN
NUL
GND
LATCH
LATCH
LATCH
TRI STATE BUFFER
PROG. OP a
PROG. OP b
MIC ON/OFF
DATA
RESET
CLOCK
D0-D7
A0-A7
AEN
RD WR
ADDRESS DECODER
TO CS OF LATCHES
RD/WR LOGIC
Vcc
DATA
RESET
CLOCK
GND
EEPROM PROGRAMMING SOCKET
INPUT BUFFER
A CHOICE
B CHOICE
SELECT
LATCH
TAPE PLAYER
ATTEN.
LATCH

FIG. 1

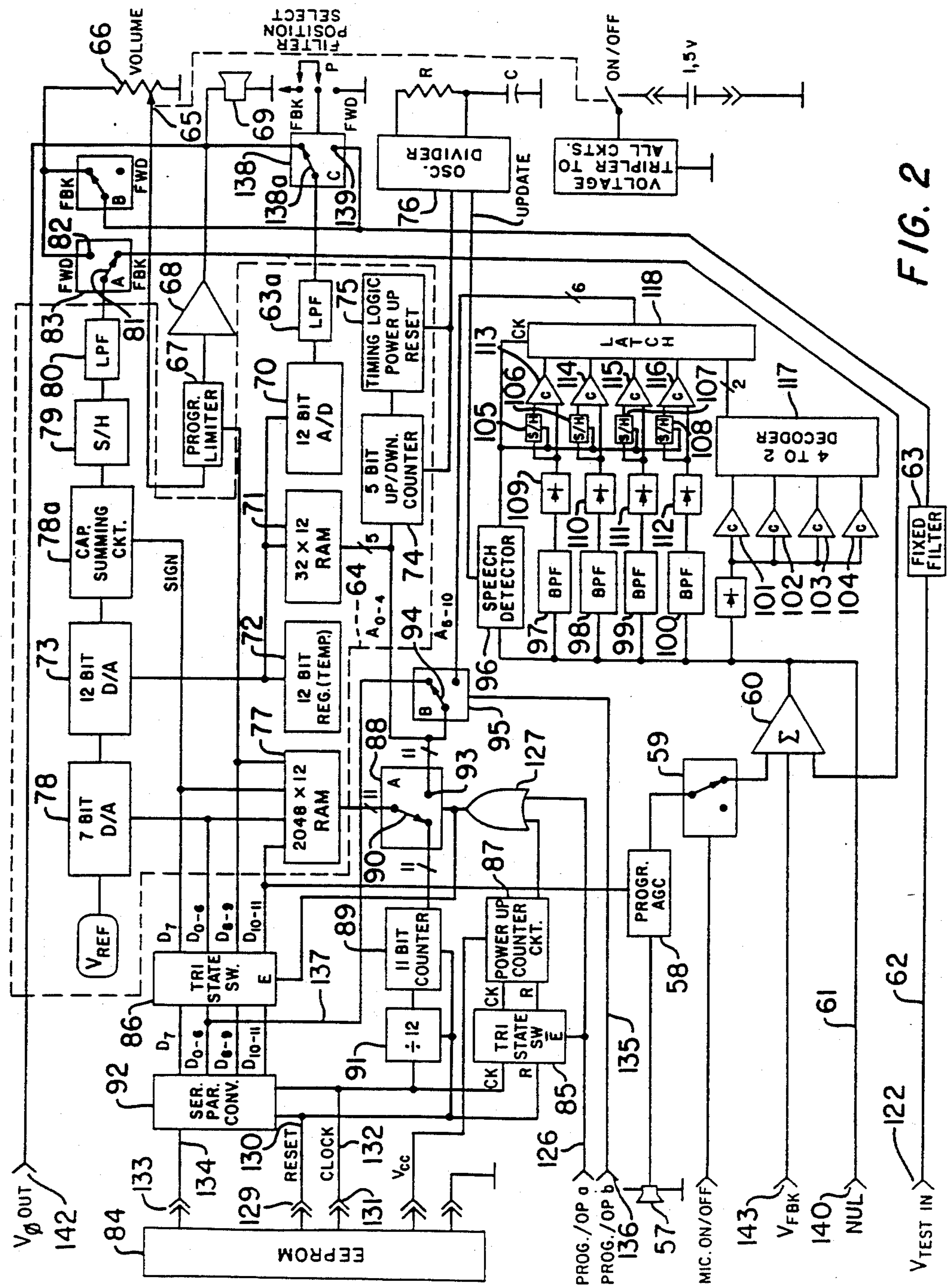
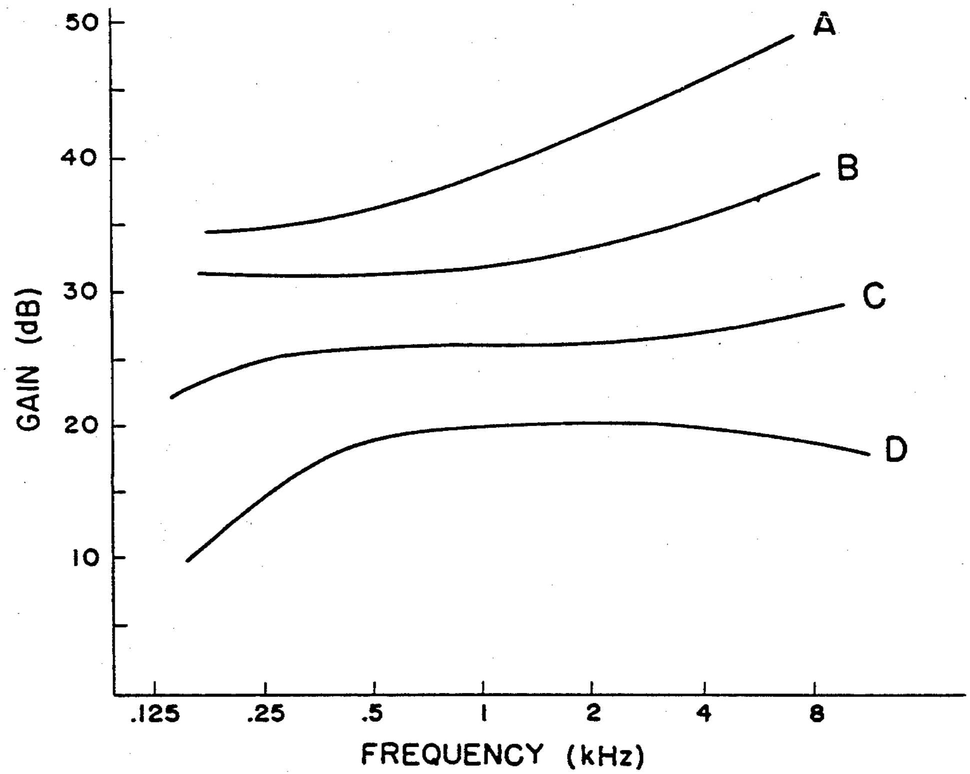
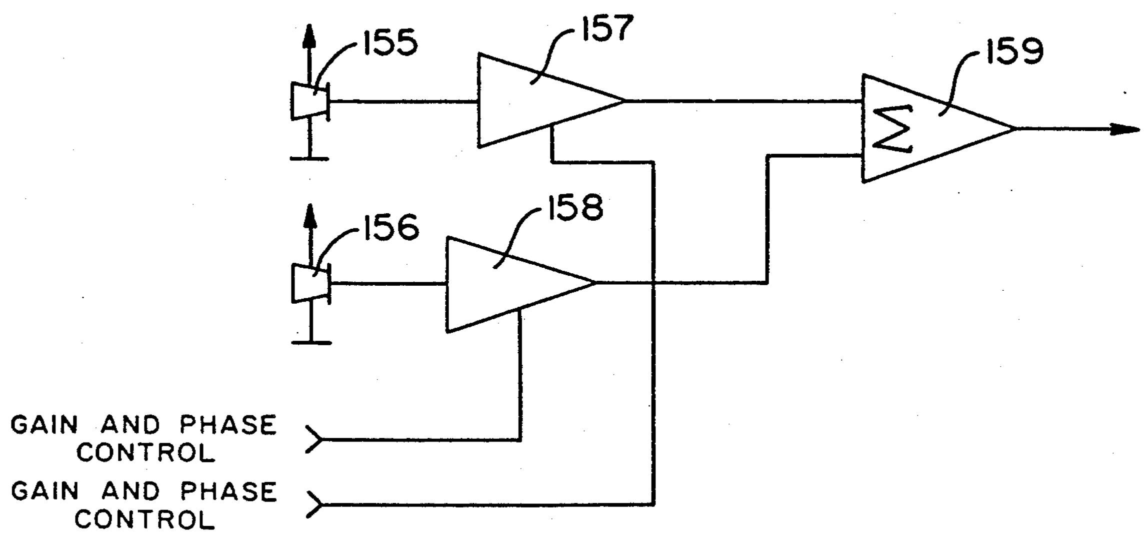

VOLUME
FILTER POSITION SELECT
ON/OFF
1,5v
OSC. DIVIDER
VOLTAGE TRIPLER TO ALL CKTS.
UPDATE
LPF
S/H
CAP. SUMMING CKT.
12 BIT D/A
7 BIT D/A
V_{REF}
PROGR. LIMITER
12 BIT A/D
32 x 12 RAM
12 BIT REG.(TEMP.)
2048 x 12 RAM
TIMING LOGIC POWER UP RESET
5 BIT UP/DWN. COUNTER
LATCH
DECODER 4 TO 2
SPEECH DETECTOR
BPF
FIXED FILTER
PROGR. AGC
TRI STATE SW.
SER. PAR. CONV.
11 BIT COUNTER
÷ 12
POWER UP COUNTER CKT.
EEPROM
SIGN
RESET
CLOCK
V_{CC}
V_{ϕ} OUT
PROG./OP a
PROG./OP b
MIC. ON/OFF
V_{FBK}
NUL
V_{TEST} IN

FIG. 2



50
40
30
20
10
GAIN (dB)
A
B
C
D
.125
.25
.5
1
2
4
8
FREQUENCY (kHz)

FIG. 3



155
157
159
156
158
GAIN AND PHASE CONTROL
GAIN AND PHASE CONTROL

FIG. 5

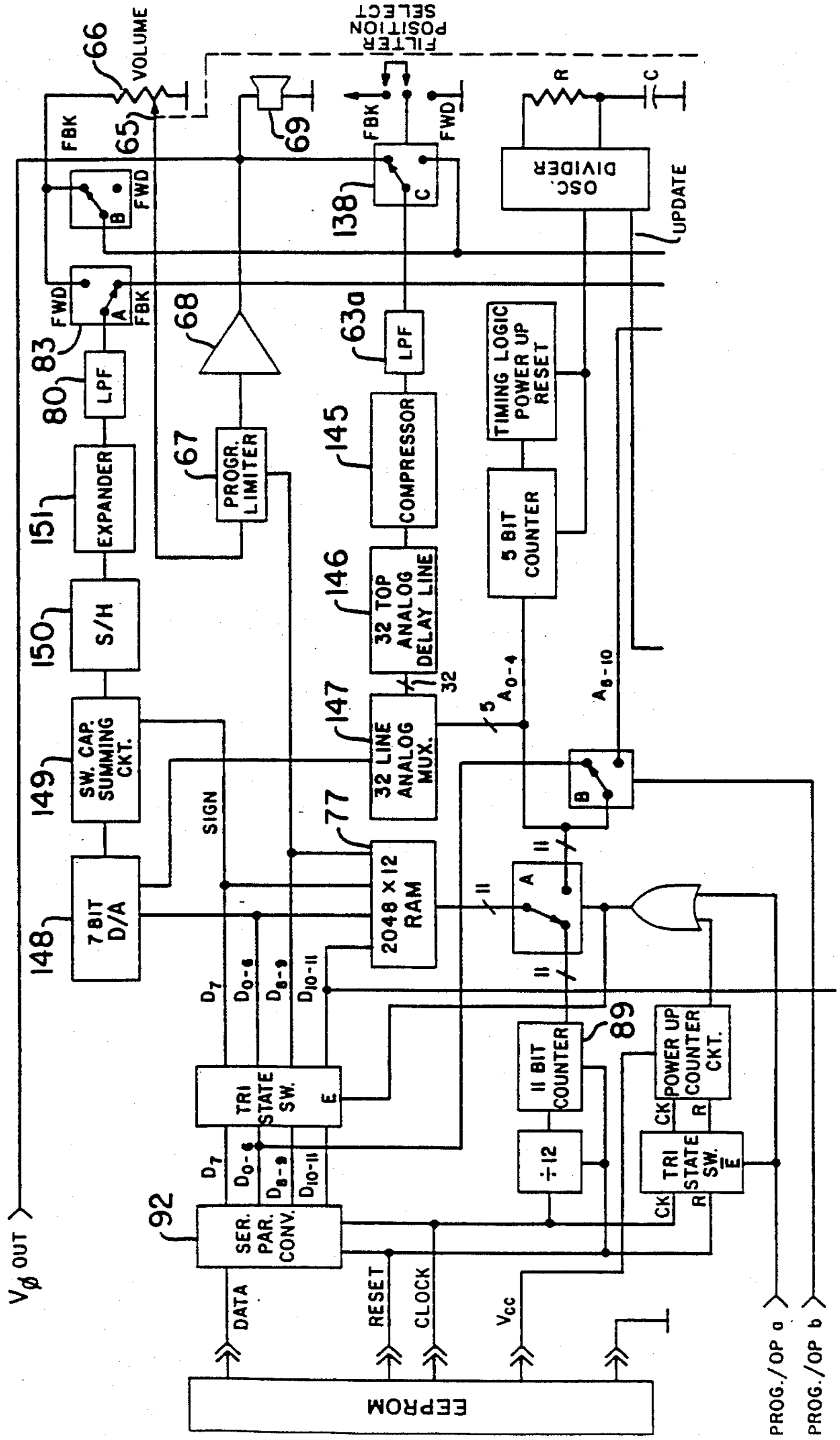

VOLUME
FILTER POSITION SELECT
FBK
FWD
OSC. DIVIDER
UPDATE
LPF
EXPANDER
S/H
SW. CAP. SUMMING CKT.
7 BIT D/A
PROGR. LIMITER
COMPRESSOR
32 TOP ANALOG DELAY LINE
32 LINE ANALOG MUX.
2048 X 12 RAM
TIMING LOGIC POWER UP RESET
5 BIT COUNTER
11 BIT COUNTER
÷12
POWER UP COUNTER CKT.
TRI STATE SW.
SER. PAR. CONV.
SIGN
DATA
RESET
CLOCK
EEPROM
PROG./OP a
PROG./OP b

FIG. 4

HOST CONTROLLER FOR PROGRAMMABLE DIGITAL HEARING AID SYSTEM

This application is a division of application Ser. No. 879,214, filed on June 26, 1986 now U.S. Pat. No. 4,731,850.

This invention relates to hearing aids, and more particularly to hearing aids that are programmable so as to have suitable characteristics to compensate for the hearing deficiencies of a patient. More specifically, it relates to hearing aids of this character that are capable of automatically adjusting to optimum parameter values as operating conditions such as speech level, room reverberation and background noise change, and also for reducing acoustic feedback.

BACKGROUND OF THE INVENTION

Conventional hearing aids suffer from several shortcomings. It is difficult if not impossible with conventional hearing aids to provide a frequency-gain characteristic that is ideal for each individual user. The acoustic coupling between the hearing aid receiver and the eardrum also introduces changes in the frequency-gain characteristic that is usually deleterious to both speech intelligibility and overall sound quality. For many patients, the optimum frequency-gain characteristic varies as a function of the level of the speech signal reaching the hearing aid. In order to protect patients from uncomfortably or dangerously loud signals, it is also necessary to limit the maximum acoustic power output of the hearing aid in some way. The methods used to limit acoustic power output of hearing aids typically introduce deleterious distortions to the amplified speech signal.

Another common problem is that of acoustic feedback. Even in the best designed hearing aids, not all of the amplified acoustic signal is delivered to the eardrum. A small proportion of the amplified acoustic signal leaks back to the hearing aid microphone forming an acoustic feedback loop. If the gain of the hearing aid is sufficiently high, this acoustic feedback will cause a self-generating oscillation to occur, resulting in an unwanted and highly unpleasant whistling sound. These acoustic oscillations prevent the hearing aid from being used. Methods of acoustic feedback control that are typically used include a tighter acoustic seal between the earmold and the walls of the ear canal so as to reduce acoustic leakage, placing the microphone at some distance from the hearing aid receiver, e.g. on the opposite ear, or simply reducing the gain of the hearing aid. None of these methods provides a satisfactory solution for high-gain hearing aids.

One of the most common complaints of hearing aid users is that background noise is particularly damaging to the understanding of speech. Methods currently used to reduce background noise in hearing aids employ filtering techniques in which the frequency regions containing high noise levels are eliminated.

Another common problem is that of room reverberation produced by acoustic reflections off the walls, ceiling, floor, and other surfaces in a room. A small amount of reverberation is beneficial but too much reverberation will make a room sound hollow or echoic and will interfere with both the quality and the intelligibility of speech.

Systems have been proposed heretofore utilizing computers for testing the hearing of patients and generating programming for a programmable hearing aid, as disclosed in "Computer Application in Audiology and Rehabilitation of the Hearing Impaired" by Harry Levitt, *Journal of Communication* Disorders 13 (1980), pages 471–481, and in the U.S. Pat. Nos. 4,187,413 to Moser, 4,489,610 to Slavin, and 4,548,082 to Engebretson et al., for example. None of these systems, however, affords a satisfactory solution for the problems of acoustic feedback, background noise, room reverberation and changes in the optimum frequency-gain characteristic resulting from variations in the level of the speech signal reaching the hearing aid.

It is an object of the invention, accordingly, to provide a new and improved hearing aid system that is free of the above-noted deficiencies of the prior art.

Another object of the invention is to provide new and improved hearing aid apparatus of the above character which is capable of automatically adjusting to an optimum set of parameter values as the speech level and type of background noise change.

A further object of the invention is to provide new and improved hearing aid apparatus of the above character in which acoustic feedback is substantially reduced.

Still another object of the invention is to provide new and improved hearing aid apparatus that is capable of effective noise and reverberation suppression and acoustic feedback reduction while maintaining optimum hearing characteristics as the speech and noise levels vary.

SUMMARY OF THE INVENTION

A hearing aid system according to the invention comprises a hearing aid that is programmable so as to have optimum electro-acoustic characteristics for the patient and acoustic environment in which it is used. It also includes instrumentation for measuring relevant audiological characteristics of the patient, as well as techniques and instrumentation for programming the hearing aid to have selected characteristics to compensate for hearing deficiencies determined from the measurements made. Desirably, several sets of optimum hearing aid parameter values, specified in terms of both the amplitude and phase characteristics, are determined for the patient as a function of speech level, type of background noise, and room reverberation, both spectral and temporal characteristics of the noise being taken into account. The selected optimum parameter values are preferably programmed into an electronically erasable, programmable read only memory (EEPROM) which supplies coefficients to programmable filter and amplitude limiting means in the hearing aid so as to cause the hearing aid to adjust automatically to the optimum set of parameter values for the speech level, room reverberation, and type of background noise then obtaining.

In one form, the programmable filter may be a digital equivalent of a tapped delay line in which each delayed sample is multiplied by a weighting coefficient. The sum of the weighted samples generates the desired electroacoustic characteristics. Alternatively, the programmable filter may be a tapped analog delay line in which the sum of the weighted outputs of the taps generates the desired characteristics. An important advantage of the latter type of filter is that the power consumption is low and quasidigital techniques can be used, i.e., the waveform is sampled at discrete intervals in time without analog-to-digital conversion.

Another form of filter uses a small number of delays in which the delayed output is multiplied by a coefficient and added to the filter input so as to achieve additional delays, a technique known as recursive filtering.

The invention also provides means for reducing acoustical feedback. In one embodiment, an electrical feedback path in the hearing aid is matched in both amplitude and phase to the acoustic feedback path and the two feedback signals are subtracted so as to cancel each other. In an alternative embodiment, a single filter in the forward path is used with a transmission characteristic equivalent to that of the filter in the forward path plus the electrical feedback path.

Environmental noise control is effected according to the invention by providing means for sensing the relative speech/noise content in the signals from the hearing aid microphone and generating binary words that are supplied to the programmable filter for selecting from a memory a set of delay line tap coefficients that are effective to impart to the filter the appropriate frequency response for the specific environmental noise condition then being detected.

According to the invention, reduction in both noise and reverberation is achieved by the use of two or more microphones. The output of each microphone is processed in both amplitude and phase such that the summed output of the microphones is analogous to the output of a frequency selective directional microphone.

DESCRIPTION OF THE PREFERRED EMBODIMENT

For a better understanding of the invention, reference is made to the following detailed description of a representative embodiment taken in conjunction with the accompanying drawings, in which:

FIG. 1 illustrates schematically a host controller for use in ,prescribing a wearable, programmable hearing aid according to the invention;

FIG. 2 illustrates schematically one form of programmable hearing aid according to the invention which utilizes a digital delay line filter;

FIG. 3 shows a set of optimum frequency-gain characteristics appropriate for different speech levels for a typical hearing impaired subject;

FIG. 4 is a partial schematic diagram of the hearing aid shown in FIG. 2, modified to utilize an analog delay line filter; and

FIG. 5 shows schematically how multiple microphones can be used according to the invention to reduce both noise and reverberation.

A hearing aid system according to the invention comprises generally a wearable, programmable hearing aid in which all operations are controlled by data stored in an erasable electrical programmable read only memory (EEPROM), and a host controller providing electrical signals and test sounds as necessary for measuring the residual hearing of a subject, establishing optimal hearing aid parameters for the subject (including the phase relationship between input and output) and generating control signals as necessary to program the EEPROM module to perform the desired operations in the hearing aid.

THE HOST CONTROLLER

Referring first to FIG. 1, the host controller 20 is shown as comprising an audiometric signal generator 21 including a signal generator 22, the frequency of which is controllable by a digital-to-analog (D/A) converter 23 controlled by a computer 24 through a latch 25. The output level of the signal generator is adapted to be adjusted by an attenuator 26 also controlled by the computer 24, through a latch 27. The output of the attenuator 26 is supplied to one terminal 28 of a switch 29, from which it can be transmitted through an amplifier 119 and a switch 120 to a terminal connector 121.

The host controller 20 also includes a phase measurement circuit comprising a digital phase shifter 30 which is adapted to receive the output voltage of the hearing aid through a connector 31 and a programmable gain amplifier 32, the gain of which is controllable by the computer 24 through the latch 33.

In adjusting a hearing aid for reduced feedback as described below, the input voltage developed by the hearing aid microphone in response to acoustic feedback alone when a test signal is supplied to the hearing aid is summed with an electrical feedback voltage supplied to the terminal 144 by the phase shifter 30 as an acoustical feedback cancellation voltage. The summed acoustic and electrical feedback voltages from the hearing aid are supplied from the terminal 34 through the engaged movable and fixed contacts 35 and 36 of a switch 37, a programmable gain amplifier 38, a rectifier 39 and an analog to digital (A/D) converter 40 to the computer 24. In effecting the adjustment, the gain of the amplifier 32 and the phase shifter 30 are adjusted until a null output from the A/D converter 40 is read by the computer 24, indicating cancellation of the electrical and acoustic signals. The settings of the phase shifter 30 and the amplifier are then used to program an EEPROM in the hearing aid so as to cancel acoustic feedback.

The programmable gain amplifier 38 is controlled by the computer 24 through a latch 41, and the digital phase shifter 30 is also controlled by the computer 24 through the latches 42 and 43.

Additional components of the host controller 20 are programming logic comprising the latches 44 and 45 and a conventional tri-state buffer 46 which are activated in response to the computer 24 to provide the necessary signals to program the EEPROM and control the hearing aid during testing and programming, as described in greater detail below.

A conventional tape player 47 is also provided for generating various sound combinations for testing of a patient's hearing. The tape player 47 is controlled by the computer 24 through a latch 48 and its output level is adapted to be controlled by an attenuator 49, also controlled by the computer 24, through a latch 50. The output of the attenuator 49 is supplied to the movable contact 51 of a switch 52 which is controllable to supply the output either to a connector 53 to a sound field system, or to a fixed contact 54 on the switch 29 for supply to the hearing aid, as described in greater detail below. The switches 29 and 52 and the tri-state buffer 46 are adapted to be controlled by the computer 24 through the latch 44.

The respective latches 27, 25, 42, 43, 33, 40, 41, 44, 45, 48 and 50 are adapted to be controlled by the computer 24 through an address decoder 55 and Read/Write logic 56.

THE HEARING AID

Referring now to FIG. 2, a wearable hearing aid according to the invention comprises a microphone 57, the output of which is fed through a programmable automatic gain control (AGC) circuit 58 and a switch

59 to one terminal of summing amplifier 60. In normal operation, amplifier 60 supplies the signal through the conductors 61 and 62 and a filter 63 to a programmable filter 64 which is adapted to be programmed in the manner described below to produce optimum hearing aid characteristics for the patient based on the measurements made of the patient's residual hearing. The output of the programmable filter 64 is fed from the movable contact 65 of the volume control 66, through a programmable limiter 67, and an amplifier 68 to the hearing aid receiver 69.

The programmable filter 64 comprises essentially a digital tapped delay line including a 12 bit A/D converter 70, which supplies outputs to a 32×12 random access memory (RAM) 71, a 12 bit temporary register 72, nd a 12 bit D/A converter 73. The signal supplied to the RAM 71 is sampled by a 5 bit up-down counter 74 controlled by timing logic 75 connected to receive clock signals from an oscillator divider 76 oscillating at a frequency at least twice the audio signal band width.

The characteristics of the programmable filter 64 are determined by coefficients stored in a random access memory (RAM) 77 which are selected and fed to a 7 bit D/A converter 78, the output of which is supplied to the D/A converter 73 for multiplication of the sampled data by a selected coefficient from the RAM 77. The output of the D/A converter 73 is summed in a conventional charge transfer summing circuit 78*a* and the sum signal is supplied through a sample and hold circuit 79 to a conventional anti-imaging filter 80. The output of the filter 80 is fed through the movable contact 81 of a switch 83 and a fixed contact 82 thereof to the volume control 66 and eventually to the receiver 69.

The coefficients stored in the RAM 77 when the hearing aid is in operation are provided essentially by an EEPROM 84 previously programmed by the host controller 20 (FIG. 1) as described in greater detail hereinafter. On power-up, the filter coefficients and limit parameters are transferred from the EEPROM 84 to the RAM 77 as follows: Tri-state switches 85 and 86 are enabled and power is supplied to the EEPROM 84 from a power-up control circuit 87. A switch 88 is now activated to connect an 11 bit counter 89 to the RAM 77 through the switch movable contact 90. The power-up control circuit 87 acting through the tri-state switch 85 supplies reset pulses to a divide-by-12 counter 91, to the 11 bit counter 89 and to a series parallel converter 92.

The power-up control circuit 87 also supplies clock pulses to the series parallel converter 92 and to the divide-by-12 counter 91 such that after every twelfth clock pulse, data is transferred from the EEPROM 84 through the series parallel converter 92 and the tri-state switch 86 to one of the memory locations in the RAM 77 as determined by the 11 bit counter 89. These steps are repeated as many times as are required to transfer all of the data stored in the EEPROM 84 to the RAM 77. When that occurs, the tri-state switch 86 is disabled and supplies a signal to the switch 88 to connect the movable contact 90 thereof to the fixed contact 93, which is connected to the movable contact 94 of a switch 95 and to the 5 bit counter 74 and the RAM 71. The hearing aid is then in its normal operating mode.

Automatic adjustment of the hearing aid to take into account environmental conditions such as changes in the speech level and type of background noise obtaining at any time is effected by environmental control means comprising a speech detector 96, four band pass filters 97, 98, 99 and 100, and a level detector including four differently prebiased comparators 101, 102, 103 and 104. Typical bandwidths for the filters 97, 98, 99 and 100 might be 100–750 Hz, 750–1500 Hz, 1500–3000 Hz, and 3000 to the upper frequency limit, respectively. The speech detector 96 is conventional and may include a level detector followed by a short term averaging device. For steady state noise, the output of the level detector will be relatively constant, indicating that noise only is present. Whenever fluctuations in level are within a prescribed bandwidth typical of speech, the short term average increases, indicating that speech is present. The speech detector 96 is clocked periodically by signals from the oscillator divider 76 to provide either a zero output indicative of noise or a one output indicative of speech periodically to each of a plurality of sample and hold circuits 105, 106, 107 and 108. The outputs of the band pass filters 97, 98, 99 and 100 are rectified in the rectifiers 109, 110, 111 and 112, respectively and fed to the sample and hold circuits 105–108, respectively, the outputs of which are fed to the comparators 113, 114, 115 and 116, respectively. The outputs of the rectifiers 109–112, respectively, are also supplied directly as inputs to the comparators 113–116, respectively.

So long as there is a one output from the speech detector 96 indicating the presence of speech in the input, the instantaneous outputs of each of the band pass filters 97–100 are compared with previous values held in the sample and hold circuits 105–108, causing the comparators 113–116, respectively, to generate a binary coefficient (0,1) indicating whether or not the speech level in the associated band pass filter exceeds the noise level. At the same time, a level detector 117 responsive to the respective outputs of the comparators 101–104 generates a two bit coefficient indicating the average signal level.

The outputs of the comparators 113–116, inclusive, and of the input level detector 117 are fed to latching means 118, which provides a six bit output that is adapted to be transmitted through the switches 95 and 88 to the RAM 77. Whenever the output of the speech detector 96 indicates that speech is present, the binary outputs of the four filter level comparators 113–116, inclusive, and that of the average level detector 117 are transmitted to the RAM 77 through the control switches 95 and 88. These binary words are updated at regular intervals at a clock rate determined by the oscillator divider 76, as stated. Each of the sixty-four possible combinations of the 6 bit binary words identifies a different frequency response for the programmable filter,and a corresponding set of coefficients stored in the RAM 77 is selected, thereby automatically adjusting the hearing aid to the optimum set of parameter values as the speech level and type of background noise change.

In prescribing hearing aids with the use of the host controller, the subject is seated in a quiet room with the hearing aid inserted in his ear. The hearing aid is connected to the host controller by an electrical cable (not shown), thereby placing it directly under the control of the host controller. The prescriptive procedure usually consists of five stages:

(1) measurement of the subject's residual hearing;

(2) derivation of an appropriate set of electroacoustic characteristics of the hearing aid from such measurements, and programming the hearing aid accordingly;

(3) measuring acoustic feedback in the hearing aid;

(4) re-programming the hearing aid to minimize acoustic feedback; and

(5) paired comparison testing of possible alternative settings of the hearing aid to determine the optimal hearing aid settings for the subject.

The subject's residual hearing is measured using signals generated by the audiometric section **21** in the host controller **20** (FIG. **1**). These signals are delivered to the hearing aid via the switch **29**, an amplifier **119**, a switch **120**, a connector **121**, a connector **122** in the hearing aid, and the conductor **62**. The signals are processed by the digital programming filter and associated circuitry in the hearing aid and are delivered to the subject's ear using the hearing aid receiver and associated coupling that the subject will actually wear after the hearing aid is prescribed. This procedure eliminates the need for any corrections in going from headphone to sound field measurements. The measurements are usually obtained with narrow band stimuli (tones, warble tones, or narrow band noise) and include threshold of hearing, various loudness levels (comfortable, loud) and loudness discomfort level.

The measurements obtained on the patient's residual hearing are used in deriving the electroacoustic characteristics of the hearing aid. The measurements of loudness discomfort level are used to program the limiter **67** so that sounds amplified by the hearing aid never exceed the patient's loudness discomfort level. The measurements of auditory threshold, most comfortable loudness level, and loudness discomfort level are used to determine the frequency gain characteristics of the hearing aid.

FIG. **3** shows four frequency-gain characteristics for a typical patient. Curve A is used when the speech signal reaching the hearing aid is relatively low, as would occur when somebody speaks in a very soft voice. Under these conditions the hearing aid provides a large amount of gain, particularly in the high frequencies. This is done to ensure that the speech spectrum is placed above the patient's threshold of hearing at all frequencies.

Curve B is used when the incoming speech signal is at the low end of the comfortable loudness range for a normal hearing person. The amplification provided places the speech spectrum at the bottom of the patient's most comfortable loudness range at all frequencies.

Curve C is used when the level of the incoming speech is moderately loud for a normal hearing person. The amplification provided places the speech spectrum at the top of the patient's most comfortable loudness range for all but the lowest frequencies. Less gain is provided at the low frequencies to reduce upward spread of masking effects; e.g., weak high-frequency sounds being masked by intense low frequency sounds.

Curve D is used when the signals reaching the hearing aid are very loud for a normal hearing person. Under these conditions the hearing aid provides relatively little gain with a significant roll off in the low frequency region in order to substantially reduce upward spread of masking effects.

A set of coefficients is derived for each of these frequency gain characteristics. These coefficients are derived using procedures that are well known in the field of digital signal filtering and are used to program filter **64** so that the hearing aid produces the required frequency-gain characteristic. The filter coefficients are stored in the RAM **77**.

The level of the incoming speech signal is determined by the level detectors **101**, **102**, **103** and **104**. The decoder **117** generates a binary word depending on the outputs of these level detectors. This binary word is transmitted to the RAM **77**, in order to select the appropriate set of filter coefficients. If the signal reaching the hearing aid consists of speech plus noise, as determined by the speech detector **96**, then alternative frequency-gain characteristics are used. These frequency-gain characteristics are derived by first determining the incoming signal level, as described above, selecting an appropriate frequency-gain characteristic and then reducing the gain in those frequency regions where the background noise level exceeds the speech level. This is determined by comparing the output levels of the bandpass filters **97**, **98**, **99**, and **100** when speech is present to the corresponding levels measured when noise only is present. The latter information is stored in the sample and hold units **105**, **106**, **107** and **108**. The comparisons are done by means of the comparators **113**, **114**, **115**, and **116**. The outputs of these comparators in combination with the outputs of the level decoder, **117**, generate a 6-bit word that selects the appropriate set of filter coefficients in the RAM **77**.

For patient hearing parameter selection and programming, the hearing aid is interfaced with the host controller as described above, and the EEPROM **84** (FIG. **2**) is plugged into a programming slot **124** in the host controller **20**. A conductor in the line **125** (FIG. **1**) is set to logic **1** by operation of the computer **24** which applies a logic **1** signal to the line **126** (FIG. **2**), resulting in the tri-state switch **85** being disabled and the switch **88** being activated by an OR gate **127** to move the switch movable contact **90** to connect the counter **89** to the RAM **77**. The host controller **20** generates reset pulses which are transmitted over a conductor in the line **128** (FIG. **1**) through the connector **129** and the line **130** in the hearing aid (FIG. **2**) to reset the counters **91** and **89** and the series-to-parallel converter **92**.

Clock signals are fed from the host controller **20** over another conductor in the line **128** through a connector **131** and a conductor **132** in the hearing aid to the serial-to-parallel converter **92** and are divided by **12** in the counter **91**, the output of which is fed through the switch **88** to the RAM **77**. Sychronously with the clock signals on the line **132**, data are fed from a conductor in the line **128** in the host controller **20** and through a connector **133** and a conductor **134** in the hearing aid to the serial-to-parallel converter **92**. After every twelfth clock signal, data is transferred from the serial-to-parallel converter **92** through the tri-state switch **86** to one of the memory locations in the RAM **77** determined by the counter **89**. This step may be repeated as many times as required to store essential data in the RAM **77** within the storage limits of the latter. The line **126** is then set to logic 0 by operation of the computer **24** through a conductor in the line **125**, thus disabling the tristate switch **86** and actuating the switch **88** to cause the movable contact **90** thereof to move into engagement with the fixed contact **93** resulting in the connection of the RAM **77** to the counter **74** and to the RAM **71**. The hearing aid is now ready for patient hearing parameter selection and programming.

The selection of the desired hearing aid parameters is accomplished by actuating the computer **24** to set the line **135** in the hearing aid (FIG. **2**) to logic 1 through a conductor in the cable **128** and a connector **136** on the hearing aid. This activates the switch **95** and supplies

data in the form of 6 bit words from the series-to-parallel converter **92** over the line **137** through the movable contact **94** of the switch **95** and the movable contact **90** of the switch **88** to the RAM **77**. The series-to-parallel converter **92** is reset by a signal from the computer **24** (FIG. **1**) transmitted through a conductor in the line **128** and the connector **129** and conductor **130** in the hearing aid, and a 6 bit word is fed into the converter **92** and placed on seven of the address lines of the RAM **77**, selecting one of sixty-four possible sets of coefficients for use in the programmable filter in the hearing aid. The selection proceeds in the like fashion throughout the process of frequency shaping of the filter.

Feedback cancellation is achieved in the hearing aid by first measuring the acoustic feed back in situ and then creating an electronic feedback path with identical amplitude and phase characteristics. The outputs of the two feedback paths are then subtracted, thereby canceling any feedback signals that might occur.

Acoustic feedback is measured with the hearing aid in the ear as it would normally be worn. An electrical test signal is applied to the terminal **122** (FIG. **2**) from the host controller **20**. A portion of this amplified acoustic signal will leak through the ear mold and reach the microphone **57**, which will then convert the signal back to electrical form and return it to the hearing aid amplifier. Feedback will occur if the total gain in the loop, i.e., from the input to the filter **63** through the filter and amplifier of the hearing aid to the output transducer **69** and from the microphone **57** back to the input to the filter **63**, exceeds unity.

For the purpose of this measurement, the feedback loop is broken between **140** and **122**. Terminal **140** of the hearing aid is then connected to terminal **34** of the host controller, terminal **142** of the hearing aid is connected to terminal **31** of the host controller, and terminal **143** of the hearing aid is connected to terminal **144** of the host controller. The programmable phase shifter **30** and programmable amplifier **32** are then adjusted by the computer **24** so as to minimize the sum of the acoustic and electrical feedback signals of the output of summing amplifier **60**.

From the settings obtained with the phase shifter **30** and amplifier **32** at cancellation (at the "null"), the host controller calculates a set of coefficients for a programmable filter to be inserted in the electrical feedback path between terminals **142** and **143** so as to cancel the acoustic feedback.

If desired, the additional programmable filter in the feedback path of the hearing aid can be eliminated by calculating the effective gain of both the forward and feedback paths and modifying the programmable filter **64** so as to include this additional gain. This implementation necessarily requires an adjustment to both the amplitude and phase characteristics of the filter **64**. Alternatively, filter **64** can be placed in the feedback path with appropriate changes in the coefficients.

The setting of the hearing aid is then checked using a paired comparison technique to determine if optimum frequency response characteristics for the patient have been obtained. This involves the provision in the host controller of one or more additional programmable filters adjusted to have frequency responses that differ systematically from that prescribed using the above procedure. In the practice of this technique, the patient compares the prescribed hearing aid against an alternative hearing aid by replacing the prescribed filter and limiter with components having alternative electroacoustic characteristics by actuation of a select switch **123** in the host controller **20**. The patient listens to speech through the hearing aid, switching back and forth two sets of electroacoustic characteristics at will by means of the switches **123A** and **123B**, choosing the characteristic which is more intelligible or preferable in some way. Paired comparisons made in this manner may be repeated with systematic deviations from the prescribed frequency response in order to determine whether another frequency response would be more intelligible or preferable. If a better response is found, the paired comparison procedure may be repeated iteratively until the optimum set of electroacoustic characteristics is found. Adaptive strategies for finding the optimum electro-acoustic characteristics are described in Levitt et al., *Journal of Rehabilitation Research and Development,* Vol. 23, No. 1, 1986, pages 79–87.

In normal operation, the EEPROM **84**, with the optimum coefficients determined as described above stored therein, is plugged into the hearing aid, and when power is turned on the coefficients are transferred from the EEPROM **84** into the RAM **77**. From then on the EEPROM is inactive in order to save power. The operating sequence is as follows: On power-up the tri-state switches **85** and **86** are enabled, power is supplied to the EEPROM **84** from the power-up control circuit **87**, and the switch **88** is activated to connect the counter **89** to the RAM **77**. The power-up circuit control **87** supplies reset pulses to the counters **89** and **91** and to the series-to-parallel converter **92**, and clock pulses to the counter **91**, the converter **92** and the EEPROM **84**. After every twelfth clock pulse, data is transferred from the series-to-parallel converter **92** through the tri-state switch **86** to one of the memory locations in the RAM **77** determined by the counter **89**. This step is repeated until all the data stored in the EEPROM **84** have been transferred to the RAM **77**. The tri-state switch **86** is then disabled and the switch **88** is connected to the switch **37**, the counter **74** and the RAM **71**.

The hearing aid is now in its normal operating mode and speech detected by the microphone **57** is amplified in the programmable automatic gain control circuit **58** and transmitted through the amplifier **60**, the filter **63** and the low pass filter **63*a*** into the programmable filter circuitry.

A so-called "bucket brigade" analog delay line may be used as a programmable filter instead of the digital delay line described above, as shown in FIG. **4**. Thus, the audio signal may be fed from the filter **63*a*** through a compressor **145** to a delay line **146** having, say, thirty-two taps therealong. The delay line **146** is clocked by a 50% duty cycle signal at a frequency equal to at least twice the audio signal bandwidth. During the logic "1" period of the clock, the audio signal is supplied into the next stage of the analog delay line. On the falling edge of the clock signal, the analog signal is held in the next stage. During the logic "0" period of the clock, an analog multiplexer **147** selects one of the thirty-two taps of the delay line and feeds the signal therefrom into the input of a 7 bit D/A converter **148**. The signal is divided down in a resistance ladder in the converter **148** as determined by a 7 bit word from the RAM **77**, the address of which is specified by the counter **89**.

The divided down voltage is then given a plus or minus sign as determined by a bit from the series-parallel converter **92** and is fed to a switched capacitor summing circuit **149**. The analog multiplexer **147** proceeds to the next tap and repeats the process just described for

a total of thirty-two times. The thirty-two voltages thus derived are summed in the summing circuit **149** and transferred to a sample and hold circuit **150**, the output of which is expanded in an expander **151** and supplied through the filter **80** to the hearing a receiver **69**.

An analog tapped delay line in which the delays are achieved acoustically might also be employed in a programmable hearing aid filter according to the invention, instead of the digital and analog delay lines described above. This might be accomplished by feeding the hearing aid electrical signals to a transducer to generate sound to travel along a tube of appropriate length and having taps in the form of miniature microphones disposed along the tube. The signals at the taps would be multiplied by selected predetermined coefficients and added to produce a resultant characteristic much in the same manner as described above.

By using an array of two or more microphones on the body (e.g., along the frame of a pair of spectacles), the weighting coefficients can be chosen in the manner described above such that the weighted summed output of the microphones, with an appropriate phase shift, is equivalent to the output of a frequency selective directional microphone. This will reduce the effects of both noise and room reverberation. A typical configuration is shown in FIG. **5** as comprising the microphones **155** and **156** supplying inputs to amplifiers **157** and **158**, the gain and phase characteristics of each of which is programmable from the RAM **77** in the hearing aid. The outputs of the programmable gain amplifiers **157** and **158** are summed in a summing amplifier **159**, the output of which would be supplied to the programmable AGC device **58**.

Automatic adjustment of the frequency response of the hearing aid as a function of speech level may also be effected by placing the programmable filter in a feedback loop in series with a programmable compression amplifier. Since the degree of feedback depends on signal level, the overall frequency response also depends on signal level.

The components shown in FIGS. **2**, **4** and **5** may be incorporated in the hearing aid or part may be contained in a pocket size case to be carried in the clothing of the person wearing the hearing aid. In the latter case, the components contained in the case may be coupled to the components in the hearing aid by a conventional FM transmission link, for example.

The invention thus provides a novel and highly effective hearing aid system for use by hearing impaired patients. By utilizing a digital or analog delay line as a programmable filter in a hearing aid, it is possible to establish optimal hearing aid parameters for the patient. Moreover, by virtue of the novel means employed for effecting automatic adjustment of the programmable filter to optimum parameter values as the speech level, room reverberation and type of background noise change and for reducing acoustic feedback, a superior hearing aid system of optimum characteristics can be prescribed for hearing deficient patients.

The several embodiments described above and depicted in the drawings are intended to be only illustrative, and modifications in form and detail are possible within the scope of the following claims. For example, more than four bandwidth filters and different frequency bands might be employed. Also, the number of taps on the digital filter may be greater than **32** as in the specific embodiment described above and the A/D converter may, of course have more than 12 bit resolution.

We claim:

1. A host controller for producing data from a computer for a programmable hearing aid to cancel acoustic feedback comprising means for receiving signals from the hearing aid and measuring phase and amplitude, means for receiving signals from the hearing aid indicative of the summation of acoustic feedback and acoustic feedback cancellation signals, and means controlled by the computer for adjusting the phase and amplitude necessary to eliminate acoustic feedback and produce a null summation.

2. A host controller as set forth in claim **1** including means controlled by the computer for transmitting phase shift and amplitude data to program the hearing aid to eliminate acoustic feedback.

3. A host controller as set forth in claim **1** including a sound generator for generating test signals supplied to the hearing aid and in which the means for receiving signals from the hearing aid includes a circuit for receiving the test signals back from the hearing aid to provide reference phase and amplitude signals.

4. A host controller as set forth in claim **3** in which said circuit includes computer controlled means for adjusting the phase shift and amplitude to produce an acoustic feedback cancellation signal.

5. A host controller as set forth in claim **2** including computer controlled means for supplying a logic signal to the hearing aid for programming and a logic signal to the hearing aid to restore it for use by the patient.

6. A host controller for programming a filter interposed in a transmission channel of a hearing aid between a microphone and a receiver from a computer associated with the host controller comprising a signal generator for generating test signals supplied to the receiver, a circuit for returning an output signal from the transmission channel, a phase and amplitude measuring circuit for measuring the phase and amplitude of the output signal, said phase and amplitude circuit including adjustable phase shift and amplitude means controlled by the computer for adjusting the phase shift and amplitude necessary to eliminate acoustic feedback and transmitting an acoutical feedback cancellation signal to the hearing aid and a circuit for receiving signals from the hearing aid indicative of the summation of acoustic feedback and acoustic feedback cancellation signals to produce a null summation when the phase shift and amplitude adjustments necessary to eliminate acoustic feedback have been achieved.

* * * * *

UNITED STATES PATENT AND TRADEMARK OFFICE

CERTIFICATE OF CORRECTION

PATENT NO. : 4,879,749

DATED : Nov. 7, 1989

INVENTOR(S) : Levitt et al.

It is certified that error appears in the above-identified patent and that said Letters Patent is hereby corrected as shown below:

First page, 2nd col., 3rd line of ABSTRACT, "as" should read --are--.

Col. 11, line 5, "hearing a" should read --hearing aid--.

Col. 12, line 52, "acoutical" should read --acoustical--.

Signed and Sealed this

Sixteenth Day of October, 1990

Attest:

HARRY F. MANBECK, JR.

Attesting Officer

Commissioner of Patents and Trademarks

JS 44 (Rev. 3/99)

CIVIL COVER SHEET

The JS-44 civil cover sheet and the information contained herein neither replace nor supplement the filing and service of pleadings or other papers as required by law, except as provided by local rules of court. This form, approved by the Judicial Conference of the United States in September 1974, is required for the use of the Clerk of Court for the purpose of initiating the civil docket sheet. (SEE INSTRUCTIONS ON THE REVERSE OF THE FORM.)

I. (a) PLAINTIFFS

ENERGY TRANSPORATION GROUP, INC.

(b) County of Residence of First Listed Plaintiff *
(EXCEPT IN U.S. PLAINTIFF CASES)

* Plaintiff is a Delaware corporation

(c) Attorney's (Firm Name, Address, and Telephone Number)

John G. Day, Esquire
Ashby & Geddes
500 Delaware Avenue, 8th Floor, P.O. Box 1150
Wilmington, DE 19899 (302) 654-1888

DEFENDANTS

SIEMENS HEARING INSTRUMENTS, INC. and
SIEMENS AUDIOLOGISCHE TECHNIK GMBH

County of Residence of First Listed
(IN U.S. PLAINTIFF CASES ONLY)
NOTE: IN LAND CONDEMNATION CASES, USE THE LOCATION OF THE LAND INVOLVED.

Attorneys (If Known)

Unknown

II. BASIS OF JURISDICTION (Place an "X" in One Box Only)

☐ 1 U.S. Government Plaintiff
☒ 3 Federal Question (U.S. Government Not a Party)
☐ 2 U.S. Government Defendant
☐ 4 Diversity (Indicate Citizenship of Parties in Item III)

III. CITIZENSHIP OF PRINCIPAL PARTIES (Place an "X" in One Box for Plaintiff and One Box for Defendant)

(For Diversity Cases Only)

	PTF	DEF		PTF	DEF
Citizen of This State	☐ 1	☐ 1	Incorporated *or* Principal Place of Business In This State	☐ 4	☐ 4
Citizen of Another State	☐ 2	☐ 2	Incorporated *and* Principal Place of Business In Another State	☐ 5	☐ 5
Citizen or Subject of a Foreign Country	☐ 3	☐ 3	Foreign Nation	☐ 6	☐ 6

IV. NATURE OF SUIT (Place an "X" in One Box Only)

CONTRACT
☐ 110 Insurance
☐ 120 Marine
☐ 130 Miller Act
☐ 140 Negotiable Instrument
☐ 150 Recovery of Overpayment & Enforcement of Judgment
☐ 151 Medicare Act
☐ 152 Recovery of Defaulted Student Loans (Excl. Veterans)
☐ 153 Recovery of Overpayment of Veteran's Benefits
☐ 160 Stockholders' Suits
☐ 190 Other Contract
☐ 195 Contract Product Liability

REAL PROPERTY
☐ 210 Land Condemnation
☐ 220 Foreclosure
☐ 230 Rent Lease & Ejectment
☐ 240 Torts to Land
☐ 245 Tort Product Liability
☐ 290 All Other Real Property

TORTS

PERSONAL INJURY
☐ 310 Airplane
☐ 315 Airplane Product Liability
☐ 320 Assault, Libel & Slander
☐ 330 Federal Employers' Liability
☐ 340 Marine
☐ 345 Marine Product Liability
☐ 350 Motor Vehicle
☐ 355 Motor Vehicle Product Liability
☐ 360 Other Personal Injury

PERSONAL INJURY
☐ 362 Personal Injury— Med. Malpractice
☐ 365 Personal Injury — Product Liability
☐ 368 Asbestos Personal Injury Product Liability

PERSONAL PROPERTY
☐ 370 Other Fraud
☐ 371 Truth in Lending
☐ 380 Other Personal Property Damage
☐ 385 Property Damage Product Liability

CIVIL RIGHTS
☐ 441 Voting
☐ 442 Employment
☐ 443 Housing/ Accommodations
☐ 444 Welfare
☐ 440 Other Civil Rights

PRISONER PETITIONS
☐ 510 Motions to Vacate Sentence
Habeas Corpus:
☐ 530 General
☐ 535 Death Penalty
☐ 540 Mandamus & Other
☐ 550 Civil Rights
☐ 555 Prison Condition

FORFEITURE/PENALTY
☐ 610 Agriculture
☐ 620 Other Food & Drug
☐ 625 Drug Related Seizure of Property 21 USC
☐ 630 Liquor Laws
☐ 640 R.R. & Truck
☐ 650 Airline Regs.
☐ 660 Occupational Safety/Health
☐ 690 Other

LABOR
☐ 710 Fair Labor Standards Act
☐ 720 Labor/Mgmt. Relations
☐ 730 Labor/Mgmt.Reporting & Disclosure Act
☐ 740 Railway Labor Act
☐ 790 Other Labor Litigation
☐ 791 Empl. Ret. Inc. Security Act

BANKRUPTCY
☐ 422 Appeal 28 USC 158
☐ 423 Withdrawal 28 USC 157

PROPERTY RIGHTS
☐ 820 Copyrights
☒ 830 Patent
☐ 840 Trademark

SOCIAL SECURITY
☐ 861 HIA (1395ff)
☐ 862 Black Lung (923)
☐ 863 DIWC/DIWW (405(g))
☐ 864 SSID Title XVI
☐ 865 RSI (405(g))

FEDERAL TAX SUITS
☐ 870 Taxes (U.S. Plaintiff or Defendant)
☐ 871 IRS—Third Party 26 USC 7609

OTHER STATUTES
☐ 400 State Reapportionment
☐ 410 Antitrust
☐ 430 Banks and Banking
☐ 450 Commerce/ICC Rates/etc.
☐ 460 Deportation
☐ 470 Racketeer Influenced and Corrupt Organizations
☐ 810 Selective Service
☐ 850 Securities/Commodities/ Exchange
☐ 875 Customer Challenge 12 USC 3410
☐ 891 Agricultural Acts
☐ 892 Economic Stabilization Act
☐ 893 Environmental Matters
☐ 894 Energy Allocation Act
☐ 895 Freedom of Information Act
☐ 900Appeal of Fee Determination Under Equal Access to Justice
☐ 950 Constitutionality of State Statutes
☐ 890 Other Statutory Actions

V. ORIGIN (PLACE AN "X" IN ONE BOX ONLY)

☒ 1 Original Proceeding
☐ 2 Removed from State Court
☐ 3 Remanded from Appellate Court
☐ 4 Reinstated or Reopened
☐ 5 Transferred from another district (specify)
☐ 6 Multidistrict Litigation
☐ 7 Appeal to District Judge from Magistrate Judgment

VI. CAUSE OF ACTION

(Cite the U.S. Civil Statute under which you are filing and write brief statement of cause. Do not cite jurisdictional statutes unless diversity.)

Title 35, U.S. Code. This is an action arising under the patent laws of the United States.

VII. REQUESTED IN COMPLAINT:

☐ CHECK IF THIS IS A CLASS ACTION UNDER F.R.C.P. 23

DEMAND $ to be determined

CHECK YES only if demanded in complaint:
JURY DEMAND: ☒ Yes ☐ No

VIII. RELATED CASE(S) IF ANY

(See instructions): JUDGE Sleet DOCKET NUMBER 05-422-GMS

DATE August 28, 2007

SIGNATURE OF ATTORNEY OF RECORD

FOR OFFICE USE ONLY

RECEIPT # ________ AMOUNT ________ APPLYING IFP ________ JUDGE ________ MAG. JUDGE ________

JS 44 Reverse (Rev. 12/96)

INSTRUCTIONS FOR ATTORNEYS COMPLETING CIVIL COVER SHEET FORM JS-44

Authority For Civil Cover Sheet

The JS-44 civil cover sheet and the information contained herein neither replaces nor supplements the filings and service of pleading or other papers as required by law, except as provided by local rules of court. This form, approved by the Judicial Conference of the United States in September 1974, is required for the use of the Clerk of Court for the purpose of initiating the civil docket sheet. Consequently, a civil cover sheet is submitted to the Clerk of Court for each civil complaint filed. The attorney filing a case should complete the form as follows:

I. (a) **Plaintiffs-Defendants.** Enter names (last, first, middle initial) of plaintiff and defendant. If the plaintiff or defendant is a government agency, use only the full name or standard abbreviations. If the plaintiff or defendant is an official within a government agency, identify first the agency and then the official, giving both name and title.

(b.) County of Residence. For each civil case filed, except U.S. plaintiff cases, enter the name of the county where the first listed plaintiff resides at the time of filing. In U.S. plaintiff cases, enter the name of the county in which the first listed defendant resides at the time of filing. (NOTE: In land condemnation cases, the county of residence of the "defendant" is the location of the tract of land involved.)

(c) Attorneys. Enter the firm name, address, telephone number, and attorney of record. If there are several attorneys, list them on an attachment, noting in this section "(see attachment)".

II. **Jurisdiction**. The basis of jurisdiction is set forth under Rule 8(a), F.R.C.P., which requires that jurisdictions be shown in pleadings. Place an "X" in one of the boxes. If there is more than one basis of jurisdiction, precedence is given in the order shown below.

United States plaintiff. (1) Jurisdiction based on 28 U.S.C. 1345 and 1348. Suits by agencies and officers of the United States, are included here.

United States defendant. (2) When the plaintiff is suing the United States, its officers or agencies, place an "X" in this box.

Federal question. (3) This refers to suits under 28 U.S.C. 1331, where jurisdiction arises under the Constitution of the United States, an amendment to the Constitution, an act of Congress or a treaty of the United States. In cases where the U.S. is a party, the U.S. plaintiff or defendant code takes precedence, and box 1 or 2 should be marked.

Diversity of citizenship. (4) This refers to suits under 28 U.S.C. 1332, where parties are citizens of different states. When Box 4 is checked, the citizenship of the different parties must be checked. (See Section III below; federal question actions take precedence over diversity cases.)

III. **Residence (citizenship) of Principal Parties.** This section of the JS-44 is to be completed if diversity of citizenship was indicated above. Mark this section for each principal party.

IV. **Nature of Suit**. Place an "X" in the appropriate box. If the nature of suit cannot be determined, be sure the cause of action, in Section IV below, is sufficient to enable the deputy clerk or the statistical clerks in the Administrative Office to determine the nature of suit. If the cause fits more than one nature of suit, select the most definitive.

V. **Origin**. Place an "X" in one of the seven boxes.

Original Proceedings. (1) Cases which originate in the United States district courts.

Removed from State Court. (2) Proceedings initiated in state courts may be removed to the district courts under Title 28 U.S.C., Section 1441. When the petition for removal is granted, check this box.

Remanded from Appellate Court. (3) Check this box for cases remanded to the district court for further action. Use the date of remand as the filing date.

Reinstated or Reopened. (4) Check this box for cases reinstated or reopened in the district court. Use the reopening date as the filing date.

Transferred from Another District. (5) For cases transferred under Title 28 U.S.C. Section 1404(a) Do not use this for within district transfers or multidistrict litigation transfers.

Multidistrict Litigation. (6) Check this box when a multidistrict case is transferred into the district under authority of Title 28 U.S.C. Section 1407. When this box is checked, do not check (5) above.

Appeal to District Judge from Magistrate Judgment. (7) Check this box for an appeal from a magistrate judge's decision.

VI. **Cause of Action**. Report the civil statute directly related to the cause of action and give a brief description of the cause.

VII. **Requested in Complaint**. Class Action. Place an "X" in this box if you are filing a class action under Rule 23, F.R.Cv.P.

Demand. In this space enter the dollar amount (in thousands of dollars) being demanded or indicate other demand such as a preliminary injunction.

Jury Demand. Check the appropriate box to indicate whether or not a jury is being demanded.

VIII. **Related Cases.** This section of the JS-44 is used to reference related pending cases if any. If there are related pending cases, insert the docket numbers and the corresponding judge names for such cases.

Date and Attorney Signature. Date and sign the civil cover sheet.

AO FORM 85 RECEIPT (REV. 9/04)

United States District Court for the District of Delaware

Civil Action No. 07-523

ACKNOWLEDGMENT OF RECEIPT FOR AO FORM 85

NOTICE OF AVAILABILITY OF A UNITED STATES MAGISTRATE JUDGE TO EXERCISE JURISDICTION

I HEREBY ACKNOWLEDGE RECEIPT OF 3 COPIES OF AO FORM 85.

8/28/07
(Date forms issued)

Matthew D. Gordon
(Signature of Party or their Representative)

Matthew D. Gordon
(Printed name of Party or their Representative)

Note: Completed receipt will be filed in the Civil Action